KINEMATIC AND ELECTROMYOGRAPHIC ANALYSIS OF WHEELCHAIR FENCING

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Kinematic and Electromyographic Analysis of Wheelchair Fencing

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CERTIFICATE OF ORIGINALITY

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ABSTRACT

Wheelchair fencing is a Paralympic sport, in which fencers compete in a fixed wheelchair at a standardized distance. Without the contribution of footwork, wheelchair fencers rely on their arms and trunks to perform all the necessary techniques. Accordingly, substantial stress was placed onto the fencers' upper extremities; particularly to fencers with deprived trunk control. Currently, no research has yet to be conducted to investigate the injury incidence and risk factors of WF. As such, the first aim of this thesis was to examine and compare the injury patterns between elite able-bodied fencers (AB) and wheelchair fencers (WF); and between wheelchair fencers with- (Category A: CA) and without- (Category B: CB) active trunk control.

A 3-year prospective cohort study was performed. Monthly interviews were conducted to the AB and WF from the Hong Kong National Squad to collect the training duration, match duration and injury data. The overall injury incidence rate (3.85/1000 hours) for WF was significantly higher than AB (2.41/1000 hours), $p<0.01$. Upper extremity injuries were predominant in WF (73.8%). Lower extremity injuries were predominant in AB (69.4%). WF had higher risk than AB in sustaining minor injury (RR: 2.35; 95% CI: 1.56-3.61), muscle strain (RR: 2.16; 95% CI: 1.34-3.56), shoulder injury (RR: 13.55; 95% CI: 3.39-17.76), and elbow injury (RR: 5.90; 95% CI: 2.45-17.21). The CB fencers had higher injury incidence (4.87/1000 hours) than CA fencers (2.99/1000 hours), $p=0.02$; and higher risk of muscle strain (RR: 1.83; 95% CI: 1.04-3.28) and shoulder injury (RR: 4.97; 95% CI: 1.82-16.87). AB and WF showed distinct injury patterns. WFs with poor trunk control were more prevalent to sustain various shoulder musculoskeletal disorders.

The second aim of this thesis was to establish the repeatability of the optical tracking and surface electromyography (SEMG) measurements during the lunge attack of wheelchair fencing. Ten WFs performed lunge attack at their maximal speed to a dummy target at a standardized distance repetitively in a single session. The mean intraclass correlations ($ICC_{3,1}$) for angular displacement was 0.73-0.98 and coefficient of multiple correlation (CMC) was 0.70-0.98. CMC and ICCs for EMG measurement was 0.70-0.94 and 0.62-0.98 respectively. The results indicated that optical tracking and SEMG methods are reliable for examining the upper limb (UL) motion during lunge attacks.

The third study aimed to compare the agreement of the three-dimensional UL kinematic measurements using optical method and inertial tracking system. Thirty healthy male participants performed shoulder, elbow and wrist movements at their maximum speeds. The Vicon Motion Analysis System and the Xsens MTx sensors simultaneously captured the resulting motions. Pearson's correlation coefficients for shoulder, elbow and wrist movements were high (0.71-0.99), $p<0.01$. Joint angles as measured by the two systems lied within the 95% limits of agreement. The results

demonstrated high agreements between the two methods for rapid UL motion analysis and substantiate their use for briskly lunge action in wheelchair fencing.

The fourth thesis project aimed to compare the kinematic and EMG data between the CA and CB fencers during lunge attack at various fencing distances. Thirty world-class foil WFs (15 CA and 15 CB) performed the lunge attacks to a hitting target at four standardized distances (100%, 105%, 110% and 115% of the normalized fencing distance) in randomized orders. UL kinematic variables (i.e. angular displacement, peak linear velocity, peak angular velocity and cross-correlation coefficient) were computed. SEMG parameters (i.e. peak EMG and integrated EMG, onset and occurrence of peak EMG, and cross-correlation) of the 8 UL muscles (upper trapezius, infraspinatus, anterior-deltoid, mid-deltoid, biceps, triceps, wrist flexors and wrist extensors) were assessed. The results showed that WFs executed a typical powerful lunge attacks by rapidly flexing and abducting their shoulder to 100^0-120^0 in combination with 50^0-70^0 shoulder internal rotation. CB fencers displayed significantly lower peak horizontal and angular velocities, larger angular displacement and altered joint coordination over their shoulder and elbow joints at 110% and 115% of the normalized fencing distance. Compared to CA fencers, CB fencers exhibited a significantly earlier onset of biceps and substantial increase in their shoulder muscle activity (i.e. peak EMG and integrated EMG) at longer fencing distances. The altered kinematics and EMG patterns might represent a unique adaptive shoulder movement strategy used by CB fencers to compensate their poor trunk control as the fencing distance increased. These movement adaptations may demand a larger muscle effort and increase the stress to CB fencers' shoulders, which may lead to a higher risk of shoulder disorders in this fencer group. This exploratory study revealed differential biomechanical responses between CA and CB fencers during the lunge attack. The results provide the foundation from which to investigate the underlying mechanisms of WF injuries, and to establish injury prevention program or rehabilitation strategies specific to wheelchair fencing.

PUBLICATIONS AND CONFERENCE PRESENTATIONS ARISING FROM THE THESIS

Part of the work presented in this thesis have been published or presented in the following forums:

A. PUBLISHED PAPER

1. Chung WM, Yeung SS, Wong AYL, Lam IF, Tse PTF, Daswani D, Lee RYW. (2012) Musculoskeletal injuries in elite able-bodied and wheelchair foil fencers – a pilot study. Clinical Journal of Sports Medicine, 22(3): 278-280.

B. CONFERENCE PRESENTATION

1. **Chung WM**, Yeung SS, Wong AYL, Lam IF, Tse PTF, Lee RYW. Three-year prospective injury surveillance study of the Hong Kong elite able-bodied and disabled foil fencers. The 3rd KHASMSS Student Conference on S port Medicine, Rehabilitation and Exercise Science, 19 Jun 2010; Hong Kong; p. 12-14. **(Winner of the Best Oral Presentation)**
2. Chung WM, Yeung SS, Wong YL, Chan WW, Lee RYW. (2011) Kinematic analysis of fencing lunge action between elite able-bodied and Paralympic wheelchair fencers – a pilot study. Oral presented to the 5th World Congress on Bioengineering, 18-21 Aug 2011; Tainan, Taiwan; p.54.
3. Chung WM, Yeung SS, Pak CH, LEE RYW. (2011) Repeatability of Vicon Motion analysis system for upper limb kinematic measurement during fencing lunge action. Hong Kong Physiotherapy Association Conference, 22-23 Nov 2011; Hong Kong. In: Hong Kong Physiotherapy Journal, 2011, 29(2): 96
4. Chung WM, Yeung SS, Chan WW, LEE RYW. (2011) Validation of Vicon motion analysis system for upper limb kinematic measurement – a comparison study with inertial tracking Xsens system. Hong Kong Physiotherapy Association Conference, 22-23 Nov 2011; Hong Kong. In: Hong Kong Physiotherapy Journal, 2011, 29(2): 96 **(Winner of the Best Poster Presentation)**
5. Chung WM, Yeung SS, Wong AYL, Lee RYW. (2013) Kinematic analysis of the lunge attack action as performed by elite able-bodied fencers and the world-class wheelchair foil fencers. Student Conference on Sports Science, Rehabilitation and Medicine 2013.
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LIST OF ABBREVIATIONS

Abbreviation	Full Term
AB	Able-bodied
ANOVA	Analysis of variance
ANT	Anterior deltoid
BIC	Biceps brachii
CA	Category A
CB	Category B
CI	Confidence interval
CMC	Coefficient of multiple correlation
CTS	Carpal tunnel syndrome
Distance_100	Normalized fencing distance
E_ext	Elbow extension
E_flex	Elbow flexion
EMG	Electromyography
F_pro	Forearm pronation
F_sup	Forearm supination
ICC	Intraclass correlation coefficient
iEMG	Integrated electromyography
INF	Infraspinatus
IWFC	International Wheelchair Fencing Committee
LL	Left-lower
LU	Left-upper
MID	Mid-deltoid
MVIC	Maximal voluntary isometric contraction
RL	Right-lower
RR	Relative Risk
RU	Right-upper
SEMG	Surface electromyography
SCI	Spinal cord injury
S_abd	Shoulder abduction
S_flex	Shoulder flexion
S_irot	Shoulder internal rotation
TRI	Triceps brachii
UT	Upper trapezius
WE	Wrist extensors
WF	Wrist flexors
W_flex	Wrist flexion
W_ud	Wrist ulnar deviation
ROM	Range of motion
WF	Wheelchair fencer
2D	Two-dimensional
3D	Three-dimensional

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CHAPTER 1

Introduction

1.1 Background

Wheelchair fencing has been an official Paralympic event since its first debut in the first Paralympic games in 1960 (IPC, 2014). Using the same rules and equipment as that of the able-bodied fencers, wheelchair fencers compete on a fixed wheelchair. Without the contribution of footwork, wheelchair fencers rely solely on their arms and trunks to perform all the necessary techniques. The highly-repetitive, asymmetrical and impulsive nature of wheelchair fencing within a confined competition space may place substantial stress onto the wheelchair fencers' upper extremity. For those wheelchair fencers with deprived trunk control, the mechanical loadings on their arms may be colossal. Given the above, it is conceivable to expect a high incidence of upper limb injuries in wheelchair fencers. Existing injury surveys revealed that over 70% of the wheelchair fencers were injured during the international fencing competition event (Lam, et al., 1995; Reynolds, et al., 1994). However, these studies were confounded by the short-term, cross-sectional retrospective design. Presently, no systematic injury surveillance has yet been conducted to examine the incidence and its associated risk factors of the wheelchair fencers. The paucity of injury statistic imposes challenges in understanding the wheelchair fencing injury characteristics. Therefore, one of the aims of this thesis was to document and compare the injury patterns between elite able-bodied fencers and wheelchair fencers, and between wheelchair fencers with- (Category A) and without- (Category B) active trunk control.

According to the van Mechelen's sports injury prevention model, possible risk factors that associated with injuries should be explored after the injury prevalence and severity documentation (van Mechelan, et al., 1992). Properly identifying the possible risk factors is essential in introducing relevant injury prevention strategies to control and minimize the injuries. In wheelchair fencing, no study has been conducted to investigate the physical risk factors and the underlying injury mechanism. Consequently, current injury management for wheelchair fencers is mainly based on the research findings from able-bodied fencing, causing serious challenge in the specificity and practicality when apply to wheelchair fencing. There is a p ressing need to elucidate the risk factors in wheelchair fencing.

Although speculative, the possible mechanism of injury in wheelchair fencing may be related to the truncated kinetic chain. Groppel (1992) stated that different body parts can be visualized as a system of chain links. The force generated by one part of the body will be successively transferred to the other body parts. The sequential activation of the kinetic chain is usually initiated from the ground where the lower extremities of the body create a ground reaction force. The sequential activation proceeds from the legs, through the hips, trunk, the scapulothoracic and glenohumeral joints, eventually reach the distal part of the arm. Any alternation in the movement patterns that do not properly activate all portions of the kinetic link system can increase the risk of injury and affect the overall performance (Groppel, 1992; Kibler, 1994). In wheelchair fencing, the eliminated footwork impairs the movement sequence of fencing motion. Further hindering of upper limb movements would be expected among highly-disabled fencers who have compromised trunk control. In order to generate sufficient attacking speed, wheelchair fencers may need to exert a higher upper limb muscular effort and/or adopt a different movement pattern. This adaptive motor pattern may inevitably predispose the fencers to a higher risk of upper limb strain and injuries. Indeed, previous studies had demonstrated that an elimination of footwork would result in substantial reduction of upper limb joint speed (Reid, et al., 2007), increase in joint range (Nunome, et al., 2002), higher in muscle activity level (Dubowsky, et al., 2009) and alteration of motor patterns (Mulroy, et al., 2004) during wheelchair basketball shooting and wheelchair propulsion. Such phenomenon is proven to be aggravated in individuals with lower trunk function when they performed the forward-reaching task (Chen, et al., 2003; Potten, et al., 1999). Given the above, the second steps following the investigation of prevalence of wheelchair fencing injury is to examine the biomechanical risk factors that may predispose wheelchair fencers to fencing related injuries. Specifically, the biomechanics of wheelchair fencing among wheelchair fencers with different trunk control abilities should be examined. As such, another aim of the current thesis was to investigate and compare the upper limb motion and motor characteristics between category A (with trunk control) and Category B (without trunk control) during the lunge attack motion (the most important and commonly-used technique to score in wheelchair fencing).

Kinematic and electromyographic studies are commonly used in sports medicine to quantify physical risk factors and have been well applied to study able-bodied fencing (Frere, et al., 2008; Frere, et al., 2011; Morris, et al., 2011). Upper limb kinematics analyses had been conducted to investigate javelin throw (Chow, et al., 2003), wheelchair tennis (Reid, et al., 2007) and basketball (Nunome, et al., 2002; Malone, et al., 2002). The movement characteristics such as speed, force, and repetition from these studies, although comprehensive, cannot be generalized to wheelchair fencing. The only research study on wheelchair fencing in literature provided some preliminary data and feasible protocol for assessing the trunk kinematics (Fung, et al., 2013). Therefore, it is necessary to examine the kinematic and electromyographic characteristics of fencing actions and to compare these between Category A and Category B wheelchair fencers so as to investigate the relative risk exposure.

The application of optical tracking system to analyze upper limb motion is relatively new, no reliability and validity data was available for the fast and briskly movement of wheelchair fencing. The reliability and validity of the optical and EMG methods had to be established before applying these technologies to quantify the physical factors associated with fencing actions.

1.2 Organization of the thesis

This thesis comprises of seven chapters. Chapter one describes the research gap and rationale for conducting injury surveillance, establishing validity and reliability for optical tracking and surface EMG methods, and investigating the kinematic and electromyographic characteristics of the Category A and Category B fencers. Chapter two summarizes the findings of our literature reviews related to our research topics. The results of the three-year prospective injury surveillance of the Hong Kong elite wheelchair fencers are presented in Chapter three, aiming to outline the unique injury pattern of the wheelchair fencers and the difference in injury pattern between the Category A and B fencers. The methods and results of the reliability tests of the kinematic and electromyographic tools are presented in Chapter four. Chapter five outlines the validity testing of the Vicon system for rapid upper limb measurement. Chapter six presents the kinematic and electromyographic analysis results. The last chapter discusses the application and contribution of this thesis to the sporting and scientific communities.

CHAPTER 2

Literature Review

2.1 Physical disability and its impact towards the society

It is estimated that more than one billion people in the world (or about 15% of the world's population) are suffering from some forms of disability, with approximately 200 million experience considerable difficulties in functioning (World Health Organization, 2011). Disability will be an even greater concern in the coming years because its prevalence is on the rise with ageing populations, global increase in chronic health conditions and improved medical care to allow a better survival rate (WHO, 2011).

The impact of physical disability is undeniably considerable to both disabled individuals and the caregivers. The World Report on Disability (WHO, 2011) indicated that people with disabilities are generally having poor health outcomes, low self-esteem and highly dependent. The negative image of the public towards disabilities further hampers people with disabilities to be fully reintegrated into their communities, leading them to have lower education achievement, less economic participation and higher rates of poverty than peoples without disabilities (Blauwet & Willick, 2012; WHO, 2011). The exclusive reliance on informal support to people with disabilities could also lead to adverse consequences to their caregivers such as stress, isolation, and loss of socioeconomic opportunities (WHO, 2011).

Medical expense related to disability is huge. The United States National Spinal Cord Injury Statistical Center (2013) estimated the lifetime costs for a person injured at age 25 amounted to US$ 4.6 million for high tetraplegia and US$ 2.3 million for paraplegia. In Australia the lifetime costs per incident case were estimated to be AUS$5 million for a person with paraplegia and AUS$9.5 million for a person with tetraplegia (WHO & ISCS, 2013). Indirect costs, such as lost earnings, generally exceed direct costs. Prevention and early effective rehabilitation are essential to lessen the impact to people with disability, their family and the society.

As one form of rehabilitation, organized sports for people with physical disability (also known as "disabled sports") were first introduced by Dr. Ludwig Guttman during the World War II as part of the recreational rehabilitation program for war-injured veterans (DePauw & Gavron, 2005). Guttmann believed that "*by restoring activity of mind and body - by instilling self respect, self discipline, a competitive spirit and comradeship - sport develops mental attitudes that are essential for social reintegration*" (Webborn, 1999).

2.2 The benefits of sports for the disabled

The evidences of health benefits of sport activities for individual with disability are numerous and apparent. Patients with physical disability participating in sports could improve their strength, coordination and endurance (Gatts & C anp, 1979). Studies had indicated that participation in sports is effective in lowering yearly physician visits, reducing re-hospitalization and preventing long-term medical complications for wheelchair users (Curtis, et al., 1986; Stotts & Warren, 1982). The benefits are not limited to physical health. Psychologically, disabled sport participants were reported to have better confidence, leading to healthier lifestyle, and subsequently improved the quality and quantity of life (Blauwet & Willick, 2012; Groah & Lanig, 2000). Increased physical fitness in disabled athletes may also help reduce the risk of injury (Fagher &

Lexell, 2014). Sociologically, highly-performed disabled sportsman may promote the mutual understanding of the disabled and able-bodied person that increases the social support to this marginalized group (Forouzan, et al., 2013). The numerous health and social benefits associated with sports participation appeal the rehabilitation specialists, healthcare administrators and governments to incorporate sports into the rehabilitation and social integration program for people with physical disabilities (Webborn, 2012). With the growing recognition of the benefits associated with sports participation, various levels and types of organized disabled sports continues to emerge.

2.3 Disabled sports

2.3.1 Brief history and global development of sports for physically disabled

In 1948, Guttman organized a sports competition event for wheelchair athletes at Stoke Mandeville at the same time when the Olympic Games were held in London. This event became the origin of the Stoke Mandeville Games, which in turn later became the modern Paralympic Games. The word "Paralympics" emphases the concept of being a "parallel" Olympics for athletes with physical disabilities. Ever since the first Paralympic Games in Rome in 1960, the number of participating disabled athletes increased substantially. There were 330 athletes from 23 countries participated in the first Paralympic Games. The number has since grown and in 2012, more than 4,200 athletes from over 150 countries had competed in the London Paralympic Games (DePauw, 2012; IPC, 2014) (Figure 2.1). Today, the Paralympic Games is the second biggest sporting event in the world (IPC, 2014). Hundreds of international tournaments covering a wide range of disabled sport events are hosted every year around the world. It is expected the number of disabled athletes participating in various disabled sport events will continue to increase.

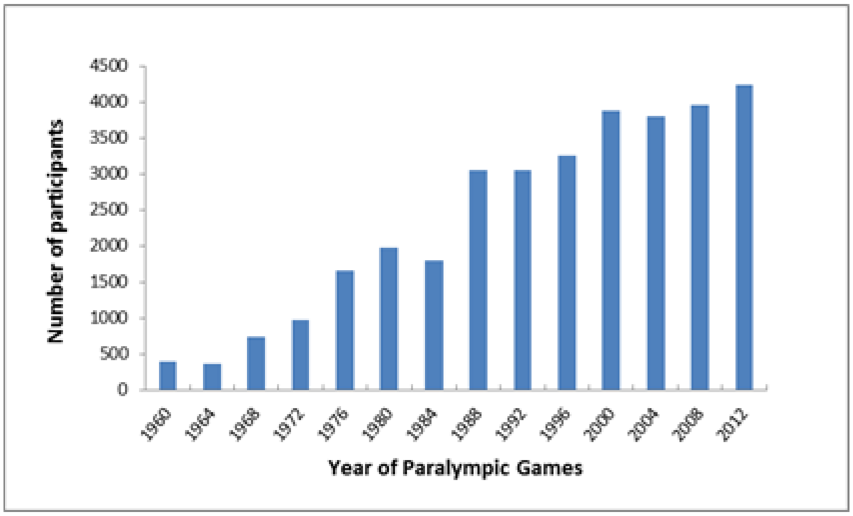


Figure 2.1 Growth of the number of disabled athletes in the past Paralympic Games (IPC, 2014)

2.3.2 The switch of disabled sports – from rehabilitation to elite competitive level

Despite the numerous benefits of sport activities, there is an inherent risk of injury while participating in sports, especially for athletes competing at elite level (Fagher & Lexell, 2014; Ljungqvist, et al., 2009; Webborn, 2012). As the original rehabilitative and recreational nature of disabled sports gradually evolved into professional and elite competitions, the intensity of trainings and competitions increase inevitably. In addition to their already higher vulnerability to musculoskeletal injuries, the increased duration and intensity of trainings and higher level of competitions inevitably increased the incidence and variety of injuries among disabled athletes (Ferrara, et al., 2000; Groah & Lanig, 2000; Webborn, 2012).

2.3.3 Sports injury in disabled sports

Table 2.1 illustrates the injury prevalence in the previous Paralympic Games. A review of the surveillance studies pertinent to the elite disabled athletes revealed that over the years, the injury rate in the Paralympic Games did not show any trend of decline (Table 2.1). For instance, the injury rate of 2012 London Paralympics Games is almost 17% higher than the 2004 Athens Paralympics Games. This probably indicated that the current intervention strategies in the prevention or rehabilitation of the injuries in elite disable athletes are not effective.

Table 2.2 presents the percent distribution of injury by body region. Upper limb injuries, with especially the shoulder and hand, are commonly reported in disabled athletes. Explicitly, wheelchair athletes tend to sustain more upper extremity injuries, particularly the shoulder joints (Fagher & Lexell, 2014). Major musculoskeletal injuries including fracture and dislocation are rare. Strains, sprain and laceration are the most common injury types (Table 2.3).

Although the reported incident rate was consistently high, the range varied hugely; from 32.0 injuries/100 athletes (Ferrara, et al., 1992) to 228.6 injuries/100 athletes (Curtis & Dillion, 1985). Such discrepancies could be stemmed from the difference in the definition of injury or the study design (Table 2.4). Most studies reported injury rates as the number of injuries per game or per player appearances, making it too generic to compare without considering the exposure time for sports and time missed after injuries. Majority of the studies examining the injuries in disabled athletes have also incorporated different types of questionnaires in form of retrospective experimental design. The possible recall bias by the sportsmen may lead to over-reporting of injuries and lower the reliability of the research studies. Prospective injury surveillance with well-defined terminology and severity of injury are warranted to collect useful information with less recall bias.

Despite the musculoskeletal problems in disabled athletes are minor and diverse in nature, the injuries could lower their sports performance and cause more serious consequences as compared to their able-bodied counter parts – particularly upper limb injuries since many of the disabled athletes solely rely on their upper extremities to performing wheelchair maneuver or day to day transfer (Pepper & Willick, 2009; Vanlandewijck & Thompson, 2011; Webborn, 2013). Due to the high incidence and severe consequence, sports injury prevention and management for disabled athletes have become contemporary issues in disabled sports medicine (Webborn & Van de Vliet, 2012).

Table 2.1 Injury prevalence in the past Paralympic Games

Study	Sport	Study Design	No of subjects	No. of injured athletes	No. of injury	Injuries/100 participants	Remarks
Burnham, et al., 1991	1998 Paralympic Games – Canadian team	R	151	124	440	71	Including illness and injury
Reynolds, et al., 1992	1988 Paralympic Games – British team	R	291	201	134	66	Including illness and injury
Nyland, et al., 2002	1996 Paralympic Games – USA team	R	304	NA	254	83.6	
Sobiecka, et al., 2005	2000 Paralympic Games – Polish team	R	114	NA	125	109.6	
Athanasopoulas, et al., 2009	2004 Paralympic Games – Paralympic village polyclinic	R	131	NA	131	100	
Willick, et al., 2013	2012 Paralympic Games -	P	3,565	539	633	117.4	

R: Retrospective; P: Prospective

Table 2.2 Percentage distribution of injury by anatomical location

Study	Sports	Head	Spine / Trunk				Upper limb					Lower limb					Others
		Head	**Neck**	**Tx**	**Lx**	**Trunk**	**Shoulder**	**Upper arm**	**Elbow**	**Forearm**	**Wrist Hand**	**Hip / Pelvis**	**Thigh**	**Knee**	**Leg**	**Ankle / Foot**	**Others**
Burnham, et al., 1991	Paralympic sports	-	5	-	8	-	**38**	-	3.6	-	11	18	-	2	6	8.3	-
McCormack, et al., 1991	Wheelchair sports	5.3	0.9	3.0	1.5	2.4	16.2	11.6	6.7	2.1	**42.3**	6.7					1.3
Reynolds, et al., 1994	Paralympic sports	-	17		11	3	9	3	-	-	12	-	-	5	-	-	**40**
Webborn & Turner, 2000	Paralympic sports	1	**33**		13	1	7	7			8	7	3	3	6	11	-
Athanasopoulas, et al., 2009	Paralympic Games	8.4		4.6	11.5	-	**27.5**	-	7.6	-	2.3	5.4	9.2	3.8	6.9	13	-
Willick, et al., 2013	Paralympic Games	2.2	5.7	2.5	5.5	3.2	**17.7**	2.1	8.8	0.8	11.4	5.2	6	7.9	5.8	12.5	2.7

Top three highest percentage of injuries were highlighted with blue color

Tx: Thoracic; Lx: Lumbar

Table 2.3 Percentage distribution of injury by type

Study	Sport	Diagnosed by	Number of subject	Injury Type											
				Sprain	Strain	Tendonitis	Bursitis	Blisters	Laceration	Fractures	Nerve injury	Head	Joint	Sore	Contusion
Curtis & Dillion (1985)	Wheelchair Sports	Self-report	128		33			18	17	5	5		5	7	
Ferrara & Davis (1990)	Wheelchair Sports	Self-report	19	48	4			6	22	6					10
McCormack & Reid (1991)	Wheelchair Sports	Self-report	90	16.3					22.5	2		2	0.6		7.8
Burhham, et al. (1991)	Summer Paralympics	Team physician / physiotherapist	275	3.7	22.2				5.5						
Ferrara & Buckley (1996)	Summer Paralympics	Self-report	128	14	60	14	9		7	8		8			
Webborn, et al. (2006)	Winter Paralympics	Team physician / physiotherapist	455	23	18	13			21	15		3	8		

Top three highest percentage of injuries were highlighted with blue color

Table 2.4 Definitions of injury and study design in existing disabled sports injury epidemiological researches

Study	Study design	No. of participants	Follow-up period	Definition of injury
Ferrara & Davis, 1990	Retrospective	19	Wheelchair (SCI) athletes participated a national training camp	Loss of practice or game participation due to injury / illness for 1 day or more
Burnham, et al., 1991	Retrospective	151	Canadian athletes participated in the 1988 Paralympic Games	Injuries / illness that were reported to the medical staff
Richter, et al., 1991	Retrospective	45	Cerebral Palsy athletes who reported an injury at 1988 Paralympic Games	Any injury / illness that was evaluated by the US medical staff during the Games
Wilson & Washington, 1993	Retrospective	83	Junior National Games	Injuries encountered during training or competition
Kegel & Malchow, 1994	Retrospective	75	Fourth and Fifth International Amputee World Soccer Cup	Any injury or illness while playing sports
Taylor & Williams, 1995	Retrospective	53	Athletes who belong to the British Wheelchair Racing Association	Pain in any part of the body that affected or prevented the athlete from training or competing for at least 1 day
Ferrara, et al., 2000	Prospective	1360	Five national or international competitions in 6 years	Any injuries or illness that was evaluated by the U.S. medical staff during study period
Nyland, et al., 2002	Prospective	304	USA Paralympians at 1996 Summer Paralympics	Any soft tissue injuries, strain, sprain, tendonitis, bursitis, contusion
Webborn, et al., 2012	Prospective	505	Athletes participated in 2010 Winter Paralympics	Any sport-related musculoskeletal complaint that caused the athlete to seek medical attention during the study period
Willick, et al., 2013	Prospective	3,565	Athletes participated in 2012 Summer Paralympics	Any sport-related musculoskeletal or neurological complaint prompting an athlete to seek medical attention, regardless of whether or not the complaint resulted in lost time from training or competition

2.4 Wheelchair Fencing

2.4.1 General features of wheelchair fencing

Wheelchair fencing was developed after the World War II and was first introduced to the International Stoke Mandeville Games in 1954. It has been an official Paralympic event since its debut in the first Paralympic games in 1960 (International Wheelchair and Amputee Sports Federation, 2014). Wheelchair fencing is a unique sport in which disabled athlete use identical weapons (foil, epee and sabre), tactics and rules as those for able-bodied fencing. The major difference for wheelchair fencing is that all fencers are required to compete in a wheelchair fastened to a frame at a standardized distance (Figure 2.2). Fencers compete on fixated wheelchair with the athletes not use their legs or rise from sitting position during the bout. The proximity of the two fencers tends to increase the pace of bouts, which also demands considerable skill.

Figure 2.2 Wheelchair fencing with foil

There are individual and team competition events in foil, epee and sabre for men; and foil and epee for women. Nowadays, over 30 nations with hundreds of wheelchair fencers are actively participating in the official and sanctioned competitions organized by the International Wheelchair Fencing Federation, the official body governing the sport of wheelchair fencing (IWAS, 2014).

2.4.2 Classification of wheelchair fencing

To be eligible to participate, wheelchair fencers must have some form of permanent disability including spinal cord injury, amputation, poliomyelitis, cerebral palsy, multiple sclerosis, muscular dystrophy and a variety of congenital disorders which do not fit into any of the traditional definition of disability. Wheelchair fencers are classified into three categories - Category A, B and C to ensure fairness and integrate athletes with different disabilities into the competition. The classification is a stringent functional classification system. It is sport-specific, based on their functional status rather than medical diagnoses, and is internationally standardized (Chung, 2008; IWFC, 2011). The classification process begins with an assessment of the athlete's disability level to determine if the minimum eligibility requirements for wheelchair fencing are met. The athletes are then required to complete a series of bench and functional tests to determine their physical functional levels. The classification can only be done by accredited international classifiers, who are either a medical doctors or physiotherapists. If there is doubt about the final classification of any individual athletes, reassessment of classification will be conducted by a panel of three experienced international classifiers.

Bench tests consist of physical examinations to ascertain the wheelchair fencers' disabilities are affecting their physical functions to what extent. There are five physical examinations included in the bench tests: muscle strength, abnormal muscle tone, joint range of motion, level of amputation and limb coordination. The bench test results provide the baseline medical information, impairment level and residual motor function of the athletes. The other part of the classification procedure is functional tests. The functional tests play an important part in the classification as athletes are required to demonstrate skills that are specific to wheelchair fencing. The classifier will extensively examine the athlete's sitting balance, arm function in executing fencing maneuvers within the wheelchair, as well as the trunk and lower limb muscle performance with or without the use of a weapon (Figure 2.3 and Table 2.5). Athletes who are unable or unwilling to participate in the classification evaluation will not be allocated a sport class and will not be permitted to compete at the respective competition (Chung, 2008; IWFC, 2011).

The combined results of bench and functional tests classify the athletes into 3 different categories. Category A fencers must have good upper limb function and sitting balance. Category B fencers have poor leg function and significant impairment in sitting balance, whereas Category C fencers have poor upper limb, lower limb and trunk control that seriously affect sitting balance (Table 2.6). It is worth noting that currently Category C was not included in the Paralympic games due to limited number of participants (Chung, 2008; IWFC, 2011).

In some occasions, an athlete may undergo re-classification. Some impairments change over time, e.g. inflective or newly-diagnosed spinal cord injury participants might have gradually improved motor control over time. In these cases, classifiers can decide if the athlete has to be re-classified again at the next competition or later (IWFC, 2011).

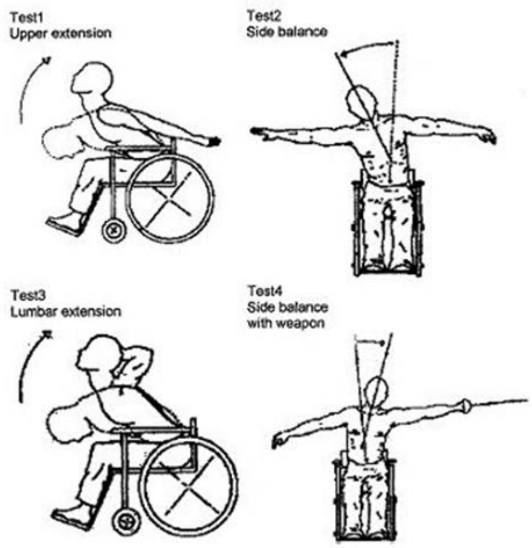


Figure 2.3 Examples of functional tests in wheelchair fencing (IWFC, 2011)

Table 2.5 Detail description of the functional test for wheelchair fencing classification (IWFC, 2011)

Functional test	Details of examination
Test 1	Consists of an evaluation of the extension of dorsal musculature: the subject, seated in the wheelchair, from a forward position of the trunk, tries to return to an upright position, contracting the dorsal muscles and maintaining the upper limbs retroflexed.
Test 2	An evaluation of lateral balance with abducted upper limbs: the athlete has to move his own center of gravity laterally to the right and left to the point where he would lose balance, thereby the lateral muscle function of the trunk and of the oblique abdominal can be evaluated as well as the lumbar muscle.
Test 3	Evaluates the extension of the trunk, but more specifically the lumbar muscles. The exercise is executed with the hands on the back of the neck, thus excluding both the inertial component of upper limb movement and the aid of the upper dorsal muscles of the trunk.
Test 4	Similar to test no 2, but presents more difficulties, since it must be executed holding the weapon, the weight of which significantly reduces the possibility of lateral inclination of the trunk without losing balance.

Table 2.6 Category, definition and examples of wheelchair fencers (Chung, 2008; IWFC, 2011)

Category	Definition	Disability groups
A	Wheelchair fencers with normal trunk and upper limb control	Lower limb amputee, low-level-lesion paraplegia (below T10), minimal involved poliomyelitis or cerebral palsy
B	Wheelchair fencers with poor trunk control and normal upper limb function	High-level-lesion paraplegia (above T10), low-level-lesion tetraplegia, incomplete-lesion tetraplegia or extensively involved poliomyelitis
C	Wheelchair fencers with poor trunk and poor upper limb control	High-level-lesion tetraplegia

2.5 Wheelchair Fencing Injury

With the footwork being eliminated, wheelchair fencers rely on their upper limb and trunk movements in order to achieve good balance, timely reactions, as well as accurate lunges and thrusts to the opponents. This asymmetrical, high velocity and repetitive upper limb and trunk movement renders fencers to different types of upper limb injuries. Unfortunately, injury studies for wheelchair fencing are rare.

For the few available injury statistics, the data were usually confined to short-term surveillance during Paralympic Games or international fencing tournaments. During the 1992 Barcelona Paralympic Games, 71% of the British wheelchair fencers were injured (Reynolds, et al., 1994). In the 2012 London Paralympic Games, a comprehensive prospective epidemiological injury survey that included 3,565 athletes from 160 delegations revealed that Paralympic wheelchair fencers had high incidence rate of 18.0 injuries/1000 athlete-days; ranking the wheelchair fencing to be the fourth highest number of injury amongst the 22 sport events. Explicitly, wheelchair fencers had the highest injury incidence amongst the various Paralympic wheelchair sport events (Willick, et al., 2013).

In the local scenario, the only available data are the one conducted by Lam and his co-workers in 1995. Lam et al. (1995) conducted a cross sectional studies to 112 disabled athletes from the Hong Kong squad team during the 10-day Far East and South Pacific Games for the Disabled. Highest incidence of injury was found in wheelchair fencing events with over 75% of the Hong Kong wheelchair fencing team members were suffering from various types of upper limb musculoskeletal injuries, particularly amongst those without active trunk control. Other than the injury incidence of wheelchair fencers, the study did not include severity, nature, and location of injury. Injury pattern and risk factors associated with wheelchair fencing injuries had not been properly addressed.

2.6 Research gap in wheelchair fencing

As one of the classical Paralympic sports, wheelchair fencing gains popularity worldwide. With the implementation of the elite training program, the number of wheelchair fencers is expected to increase. Latest injury surveillance during the 2012 London Paralympic Games showed that the prevalence of musculoskeletal injuries in wheelchair fencing was highest amongst the various Paralympic wheelchair events. The demand for proper prevention and management of wheelchair fencing related musculoskeletal injury is overwhelming. Intervention strategies in the prevention of injury are pivotal in the whole management process. To achieve this, documentation of the prevalence, severity of the injury patterns forms the first step in the prevention model (van Mechelan, et al., 1992). Indeed, according to van Mechelen's sports injury prevention model, there should be 4 s teps in the model. The initial one is the documentation of the prevalence and severity of the problem i.e. the injuries statistics. This should then be followed by identification of the possible risk factors associated with the injuries. With the proper identification of the risk factors, intervention strategies can then be formulated and introduced to the sports concerned. If the intervention strategies are effective, the prevalence and severity of the problems will then be decreased. Figure 2.4 summarizes the key steps of this model.

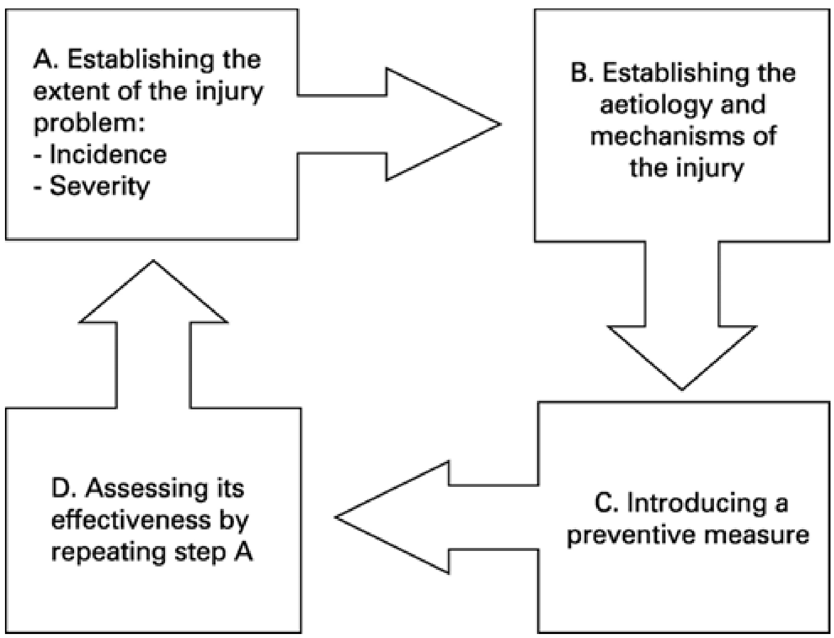
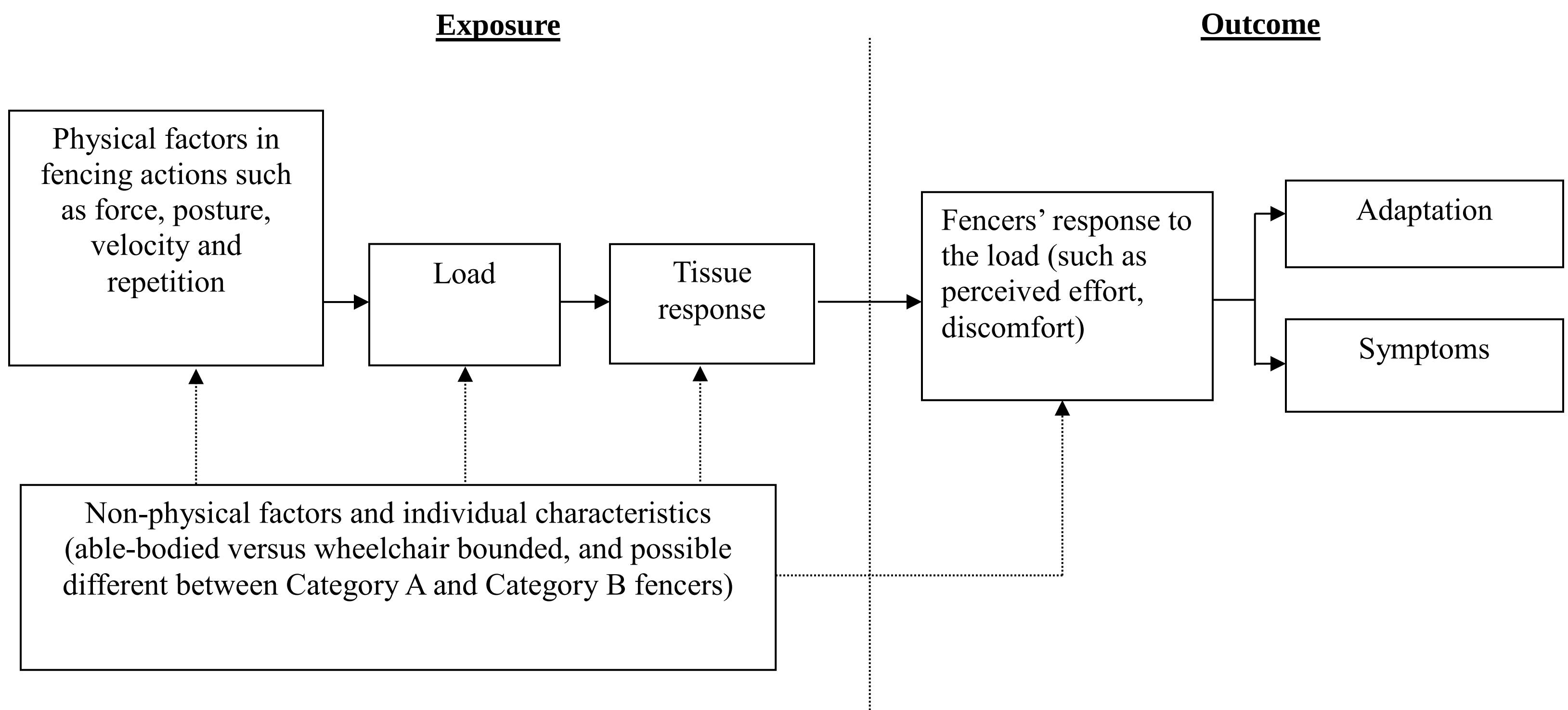


Figure 2.4 van Mechelen's recommendations for injury prevention (van Mechelen, et al., 1992)

As far as the sport of wheelchair fencing is concerned, it is obvious that the current scientific arena had not provided a clear picture on the injury patterns of this sport. The injury surveillance studies conducted in the past are predominately cross-sectional studies with usually largely varied or ill-defined injury. Possible risk factors associated with the musculoskeletal injuries for wheelchair fencing are yet to be determined. Without a proven scientific background, any injury management programs are not evidence based. In this connection, there is a need to systematically document the injuries patterns, the possible risk factors such that effective intervention strategies can be formulated, applied and tested to the wheelchair fencers. The current study proposal aims to bridge the scientific evidence of the injury patterns and the possible risk factors associated with wheelchair fencers. Since the fencing maneuvers are different between able bodied fencers; and fencers with different levels of disability, the documentation of the injury patterns and exploration of the risk factors must consider the characteristics of the fencers. Otherwise, the research findings might not be applicable or specific to the target fencer groups. For risk factors, it is well acknowledged that intrinsic and extrinsic factors constituted the two main domains of musculoskeletal injuries, and the two domains are virtually inter-related. The environment, the training/competitive facilities/equipment, the training routine, movement characteristics (physical factors) are some of the commonly investigated extrinsic risk factors associated with different kinds of sports. Among them, the biomechanical approach is commonly used to document the physical risk factors associated with the sports.

A thorough understanding of the biomechanical features in wheelchair fencing is an important step to provide useful information to design relevant injury prevention program. Figure 2.5 provides a conceptual model of the risk exposures and outcome of fencing related musculoskeletal injuries. The typical physical risk factors associated with the fencing actions are the force, posture, velocity and repetition during the fencing maneuver. These physical factors constituted corresponding load and tissue responses to the respective joint and musculoskeletal system. Such loading will be modulated by the individual's characteristics such as level of experience, and level of disability. The response of the loading under optimum or ideal situation will result in adaptation and the performance of the fencers will be enhanced. However, if the loading is excessive, symptoms or injuries might result and this is the loading that should be avoided.

Figure 2.5 A conceptual model of the risk exposures and outcome of wheelchair fencing musculoskeletal injuries



2.7 Possible physical risk factors associated with wheelchair fencing injuries

Despite the fact that there is a limited study in the identification of the physical risk factors associated with wheelchair fencing, it is speculated that the unique feature of wheelchair fencing that fencers compete in the seated position with no footwork could be considered as a cause of higher physical risk loading to the upper limbs. In higher disability group (i.e. Category B fencers) with compromised active trunk control, additional stress may put over the fencers' upper extremity and resulted in higher risk of upper extremity injury. Indeed, studies had indicated that poor lower limb and trunk control have effect on the movement and motor characteristics in a variety of activities, included wheelchair fencing (Fung, et al., 2013), other disabled sports events (Reid, et al., 2007), wheelchair propulsion (Dubowsky, et al., 2009) and functional tasks like reaching in sitting (Seelen, et al., 1997; 2001).

2.7.1 Able-bodied fencing versus wheelchair fencing

2.7.1.1 The contribution of footwork in able-bodied fencing

Fencing is an open-skilled combat sport; points are scored by touching the opponent through a weapon (Aquili, et al., 2013; Guilhem, et al., 2014; Murgu, 2006). There are three different types of weapons – foil, epee and saber, used by both able-bodied and wheelchair fencings. The gear, weapon, rules and regulations, are the same in both sports. However, the two sports have variations in their movements, possible mechanisms of injury and thus management approaches.

The success of a lunge attack is largely dictated by the speed of the execution (Turner, et al., 2013). Able bodied and wheelchair fencers achieve this execution via different techniques. In able-bodied fencing, the most important technique is the fencing lunge footwork (Suchanowski, et al., 2011). Most of the existing biomechanical analyses in able-bodied fencing were indeed directed towards the lunge attack motions (Frere, et al., 2008; Gutierrez-Davila, et al., 2013; Suchanowsbi, et al., 2011; William & Walmsley, 2000). The motor activation sequence for lunge attack starts from rear leg for propulsion towards the lunge direction, with the propel power generated from the hip abductors, knee extensors and plantar flexors (Gholipour, et al., 2008; Morris, 2011; Suchanowski, et al., 2011). This is then followed by attacking arm for weapon touch, and finally front leg for braking (Gebhardt, 1981; Morris, et al., 2011; Suchanowski, et al., 2011). With the lower limbs as the main power source, muscle activity performed by able-bodied fencer over the fencing arm is comparatively small during the lunge attack (Williams, et al., 2000; Suchanowski, et al., 2011). The motor activity sequence in upper limbs varies amongst fencers with different level of experience (Frere, et al., 2011). The results of wavelet EMG analysis by Frere, et al. (2011) showed that elbow extension phase was the major component for upper limb during lunge attack motion, while deltoid muscle was the primary mover to accelerate the humerus for flexion initiation. Playing a stabilizing role, infraspinatus generated peak activity to maintain shoulder flexion – this specialized recruitment pattern allowed infraspinatus to stabilize the shoulder. Guilhem, et al. (2014) reported the mechanical effectiveness (i.e. movement velocity) of a high-level fencing assault in able bodied fencing relies mainly on lower limbs. Hip, knee and ankle extensor muscles of the rear leg contribute to the propulsive actions, while the extensor muscles of the front leg were mainly involved during braking contractions to decelerate the body mass during the lunge action. The muscle strength imbalances between the front leg and the rear leg could result in a higher lower limb injury rate.

The asymmetrical and impulsive nature of lunge attack renders fencers to sustain a range of musculoskeletal injuries (Moyer & Konin, 1992; Roli & Bianchedi, 2008). Knee and ankle disorders including sprain and strain injuries, are common due to sudden, explosive, and ballistic movements of starting and stopping, jumping and stepping forward and backward (Harmer, 2008). Quadriceps strains at the front leg are common as the eccentric contraction is required to control the knee motions (Guilhem, et al., 2014; Murgu, 2006). Muscle imbalance is also commonly found between elite fencers' front and rear legs (Nystrom, et al., 1990; Tsolakis &d Katsikas. 2006). Studies identified that repeated execution of the task facilitates neuromuscular adaptations, causing strength asymmetries between front and rear leg muscles in the able-bodied fencers (Nystrom, et al., 1990; Roi, et al., 2008; Sapega, et al., 1984). Such muscle imbalance may increase the risk to develop lower extremity musculo-articular injury (Guilhem, et al., 2014).

2.7.1.2 Restriction of footwork in wheelchair fencing

Wheelchair fencers present a v ery different movement mechanism. While lunge attack remains to be the most important technique, the assault was completed without footwork contribution. Since the wheelchair is fixated, the spatial displacement for wheelchair fencers is very limited. Unlike the able-bodied fencers, wheelchair fencers could not step backward or forward and have to rely solely on their trunks and upper limbs for all tasks, including the powerful reach and point lunge attack. The non-fencing arm serves to maintain the sitting balance during the attack and retreat actions by holding onto the supporting bar (Fung, et al., 2013). The weight and long lever arm of the weapon may create additional torque to the fencing arm musculatures. This combination of biomechanics is unique in wheelchair fencing.

For fencers with poor trunk control (i.e. Category B), fencing motions must be performed by the upper extremity in isolation without normal trunk synergistic stabilization; a higher upper limb muscular demand is anticipated. However, the actual physical loading or biomechanics of wheelchair fencing is still unknown. Kinematic studies on a ble bodied fencing are not applicable to wheelchair fencers due to the differences.

In an attempt to explore the previous studies in the investigation of the biomechanics of wheelchair fencing, a literature search was performed for publications related to wheelchair fencing biomechanical study through MEDLINE, Sport Discus and Google Scholar databases; utilizing text words included “wheelchair fencing biomechanics", “wheelchair fencing kinematics” and “wheelchair fencing motion”. These searches identified only one article published in English between January 1966 a nd January 2014. The article by Fung, et al. (2013) was a pilot study examined the trunk kinematics of wheelchair fencers using two-dimensional optoelectrical video motion system. Maximum trunk velocity and angle in frontal plane for 14 world-class fencing (9 Category A and 5 Category B) were measured during the lunge attack and fast return actions in a classification test setting as compared to the real competition context. Both the maximum trunk velocity and angle for Category A (with trunk control) was significantly higher than Category B (without trunk control) in classification test condition only, but not during the competition condition. Fung suggested the difference could be due to the supporting bar. Wheelchair fencers are allowed to use their non-fencing arm to hold onto the supporting bar for balance during the competition context, while such support was not allowed during the functional classification test. However, the extent of the use of non-fencing arm on t he supporting bar was not quantified in this study. In fact, one key limitation of the study was the use of 2-dimensional (2D) motion analysis on complex spatial orientation of the torso. The 2D method of motion capture might confront projection errors and affect the accuracy of the kinematic results.

To standardize the distance for kinematic analysis between the classification testing and competition conditions, fencing length of lunge attack in Fung's study was set at the normalized distance (an official starting position) of the two wheelchair fencers. This fencing distance may be too short to provide sufficient challenge to the fencers' trunk control. Indeed, it may limit the generalizability of the results for real competition context.

Fung's study provided a preliminary biomechanical testing for wheelchair fencing within the laboratory setting. The upper limb kinematic characteristics that are crucial to the understanding of the possible injury mechanism in wheelchair fencers, however, had not been reported.

2.7.2 Kinematic characteristics of wheelchair sports and its implication for possible wheelchair fencing injuries

2.7.2.1 Influence of footwork elimination and compromised trunk control on upper limb kinematics

Early study by Toyoshima, et al. (1974) has quantified the substantial reduction of upper limb performance resulting from restriction in trunk and lower limb movement. By constraining the lower body and trunk of the fielders, the peak ball velocity during normal overhead throw was reduced by 36.5%. In a similar study, mathematical models were used to investigate the sequential muscle activation during throwing activities (Alexander, 1991). The results showed that systematic restriction of trunk motion could significantly reduce the maximal ball throw velocities by 50%.

Multiple studies had compared the performance of wheelchair athletes and their able-bodied counter parts, showing the similar results. The pre-impact racquet velocities in wheelchair tennis players are substantially lower in comparison to the able-bodied players (Reid, et al., 2007). Similarly, wheelchair basketball players were showed to have substantial smaller values for the vertical component of ball release velocity (4.26°/s versus 5.45°/s) and maximum wrist flexion angular velocity (878.4°/s versus 1445.9°/s) than the able-bodied basketball players in sitting position (Nunome, et al., 2002). Limited by impaired trunk control, some wheelchair basketball players failed to generate a sufficient ball release velocity for shooting as compared to able-bodied basketball players. It is conceivable that similar outcomes may be observed in wheelchair fencers. In order to generate sufficient attacking speed, these athletes rely more on upper limb to generate enough strength and velocity, thus maybe subject to higher upper limb strain. More compensation would be expected among fencers without active trunk control.

A few biomechanical studies on manual wheelchair propulsion by subjects with various level of spinal cord injury (SCI) all showed subjects with higher level SCI have larger range of motions and higher angular accelerations of their upper limbs (Boninger, et al., 2005; Davis, et al., 1998; Koontz, et al., 2002; Newsam, et al., 1999), indicating the compromised trunk movement required more upper limb movement adaptations, predisposing to a higher risk of injury, as explained by the kinetic chain concept.

2.7.2.2 Kinetic chain and injury

Kinetic chain is a concept of how different body segments are connected and function as a unit. The body can be divided into proximal and distal segments. Starting from the ground, proximal segments are comprised of the trunk and lower extremity and the distal segments are comprised of the shoulder, wrist and hand. Both segment movements are connected to form a kinetic chain. Groppel (1992) applied the kinetic link system to the analysis and description of optimal upper extremity sport biomechanics. The sequential activation of the kinetic link system originates from the ground where the lower extremities create a ground reaction force. The sequential activation proceeds from the legs, through the hips, trunk, the scapulothoracic and glenohumeral joints, eventually reaches the distal part of the arm. The kinetic chain sequence is important in generating linear momentum, angular torque and power in upper extremity sport activities, such as the throwing motion and tennis serve. In a proper energy transfer situation, each segment should start its motion at the instant of greatest speed of the preceding segment and would reach a maximum speed which is greater than that of its predecessor - this is called the summation of speed principal (Lintner, et al., 2008; Putnam, 1993). The graphical illustration of the kinetic chain is represented in Figure 2.6. The legs and trunk work sequentially in order to accelerate the shoulder for optimal upper extremity force production (Pappas, et al., 1985).

The kinetic chain principle is important in analyzing sport performance or exercise movement patterns. A typical dynamic overhand throw would include the correct sequence of stride, pelvis rotation, upper torso rotation, elbow extension, shoulder internal rotation and wrist flexion (Fleisig, et al., 1996). Any alternation in the movement patterns, either the kinetic link system is not activated in the right sequentially or part of the whole system is omitted that can increase the risk of injury and affect the overall performance (Groppel, 1992; Kibler, 1994; Lintner, et al., 2008). For example, power not generated by the proximal segment had to be compensated by the distal segment. Kibler (1998) quantified that a 20% decrease in kinetic energy delivered from the hip and trunk would necessitates a 34% increase in rotational velocity at the shoulder to deliver the same amount of force to the arm. Reid, et al. (2007) found that wheelchair tennis players have 33% lower peak pre-impact absolute and horizontal racquet velocities because of the lack of the power generated from the proximal segment of the kinetic chain. Any change in one or more components of the kinetic chain, especially in the proximal part, will alter the distal segments and cause disruption in distal segment performance, which in turn could result in excessive loading and the development of disorders over time (Bedi, 2011). Figures 2.7 and 2.8 are graphical illustration of the principle of non-optimal use of the kinetic link. Figure 2.7 shows the effect on kinetic chain when a segment is deleted from the sequential activation pattern, where Figure 2.8 shows mis-timing of the kinetic chain.

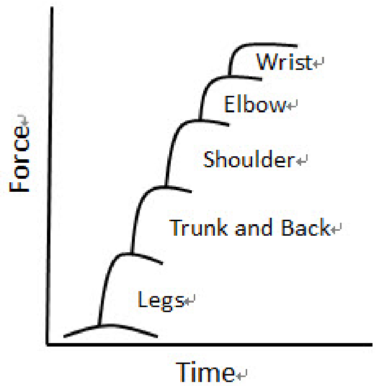


Figure 2.6. The kinetic chains of the tennis serve motion. The force starts with the ground reaction force and travels through successive links to the wrist and racquet (adopted from Kibler, 1994)

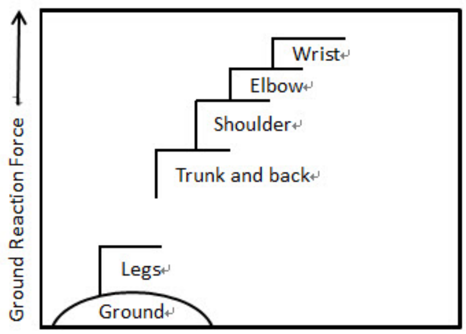


Figure 2.7 Omitting a segment from the kinetic link system (adapted from Groppel, 1992)

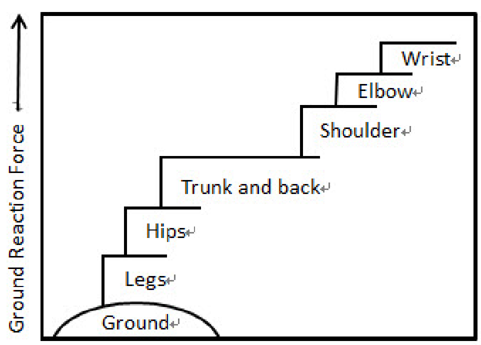


Figure 2.8 Mis-timing a link in the kinetic link system (adapted from Groppel, 1992)

2.7.3 Motor characteristics and its implication for possible wheelchair fencing injuries

2.7.3.1 Influence of footwork elimination on upper limb muscle activity

According to kinetic chain theory, a much higher upper limb muscular effort is anticipated in wheelchair fencers since their proximal segment does not exist. For Category B fencers, their trunk control is also compromised in addition to lack of footwork; an even higher upper limb effort is required. Such greater muscle activation was seen in some recent research results.

Dubowsky, et al. (2009) compared the EMG activity level between able-bodied individuals and persons with paraplegia during wheelchair propulsion and found the substantial increase in integrated EMG level over the posterior deltoid, biceps and triceps (increased by 36%, 132% and 67% respectively) in paraplegic group. A higher upper limb muscle effort in terms of peak EMG and integrated EMG level was also found in paraplegic individual with poor trunk control (e.g. injury level at T3) as compared to poor trunk control subjects (e.g. injury level as L1/L2). The authors proposed that the greater muscle effort in paraplegic individuals with truncated trunk control may result in a greater resultant shoulder and elbow joint force, which in turn leads to shoulder and elbow pathologies. Supportive findings by Louis & Gorce (2010) had revealed the higher muscle activation for upper arm, shoulder and scapular muscles in paraplegic group as compared to able-bodied group during self-paced wheelchair propulsion.

By investigating the influence of residual motor control on uppe r limb muscle activity, Mulroy, et al. (2004) examined the EMG pattern, level and duration of EMG in individuals with different spinal cord injury (SCI) levels during wheelchair propulsion. Their result indicated that the level of SCI significantly affected shoulder recruitment patterns and muscle activation level during wheelchair propulsion. Exaggerated by the denervated upper extremity muscles, both high-level (C6) and low-level tetraplegia (C7-8) exhibited a longer electromyographic duration and intensity of the shoulder muscles to propel the wheelchair as compared to high-level (T2-9) and low-level (T10-L3) paraplegia, who had normal innervation of their upper limb function. With further comparison between the high-level and low-level paraplegic groups, a slightly higher but statistically non-significant EMG activity was found in high-level paraplegia. The authors suggested that the small difference between the two groups may be related to the slow propulsion speed selected in this study. At this slow propulsion speed, the lack of trunk support in subjects with high paraplegia could be compensated and stabilized by the backrest of the test wheelchair, for which the effect of trunk control on t he shoulder muscle could not be adequately judged in the study.

2.7.3.2 Influence of truncated trunk control on upper limb muscle activity

Individuals with varying degrees of SCI experience different severity of sensorimotor impairments; depending on the level and completeness of the lesion. The motor control and coordination of the trunk and upper limb joints, as well as the strength-generating ability of the upper limb muscles, could be affected at various extents according to the severity and level of the lesion. Many studies agreed that in general the higher the lesion level, the worse the individual's sitting stability become. Gauthier, et al. (2013) have shown individuals with a higher lesion level (vertebral lesion level T7 and higher) have more limitations in multidirectional seated postural stability when compared to individual with a lower lesion level (lower than T7) and to able-bodied individuals. Chen, et al. (2003) have also shown that individuals with a higher lesion level (T6 and higher) have decreased dynamic sitting stability compared to individuals with lower lesion levels (T7 and lower). Such decreased stability is associated to the partial or complete loss of the abdominal and lower back voluntary muscle control as the levels of spinal cord lesion become higher. This loss of motor function and reduced dynamic stability could in turn affect the strength-generating ability and the muscle synergies at the upper limbs, especially when thoracohumeral muscles were involved (Chen, et al., 2003; Potten, et al., 1999). Wheelchair fencers with any abdominal and low back muscle paralysis would experience diminished dynamic stability and potential reduction in strength-generating abilities; thus altering movement strategies as well as sustaining upper limb joints load.

Gagnon, et al. (2003) investigated the upper limb muscle activities between individuals with low-level (T11 to L2) or high-level (C7 to T6) spinal cord injury during the posterior transfers performed in long sitting position. Although the patterns and magnitudes of the angular displacements were found similar between the two groups, a major difference in movement characteristics was found. The high-level spinal cord injury generally initiated the task from a forward flexed posture, whereas an almost upright alignment of trunk posture was adopted by the low-level spinal cord injury participants. Further, a significantly higher muscular demand of the upper limb muscles was found in participants with high-level spinal cord injury. Gagnon suggested the potential need of higher upper limb muscle work and alternation of movement pattern to compensate for the additional trunk and upper limb musculature impairment in individuals with poor control. Biomechanical results confirmed the influence of trunk impairment on upper limb muscle activity and movement pattern to accomplish different functional tasks.

2.7.3.3 Influence of footwork elimination on upper limb muscle recruitment pattern

Electromyographic (EMG) data showed SCI subjects with postural instability adopted compensatory use of scapula and shoulder muscles during forward reaching task in sitting position. Such shoulder muscle EMG activities significantly increased when forward-reaching distance increased beyond arm-length (Seelen, et al., 1997). This alternative motor strategy over the shoulder complex was found to be more profound in subjects with higher SCI lesions (Seelen, et al., 1997; 2001). Under normal circumstances, the primary role of muscles at the glenohumeral and scapulothoracic region is to move the upper limb. As dynamic trunk control is compromised in SCI subjects, such muscle group serves as the postural stabilizers for trunk instead; thereby renders the glenohumeral muscular imbalances and increase the risk to develop shoulder disorders (Yildirim, et al., 2010). Although both were wheelchair users, wheelchair fencers are not identical to SCI subjects in Seelen's study; the findings could not be fully transferable. Furthermore, the anatomical planes involved are different. In wheelchair fencing, the lunge attack task predominantly occurs in coronal plane whereas in Seelen's study, forward-reaching task was studied and the task mainly occurred in sagittal plane. However, it is logical to expect altered motor recruitment pattern and muscle activity of the fencing arm muscles particularly when performing lunge attack at longer fencing distance.

Deficit in optimal motor regulation and movement patterns are widely accepted to be the crucial factor for the development of chronic musculoskeletal dysfunction (Comerford & Mottram, 2001; Hodges, et al., 2003; Worsley, et al., 2013). Recent emerging evidences support the relationship between altered muscle recruitment patterns and musculoskeletal disorders. The experimental work by Cholewicki, et al. (2005) demonstrated that delayed trunk muscle response was highly associated with the onset of back and lower extremity injuries in young athletes. Muscle reflex response to sudden force release in trunk flexion, extension, and lateral flexion was prospectively measured and analyzed in 292 college athletes who were followed up to 3 years to track low back injuries. Delayed trunk muscle reflex responses were identified to be a preexisting risk factor prior to low back injury. Zazulak, et al. (2007) also associated the neuromuscular control of trunk to knee injury; the results led to a conclusion that the neuromuscular control of trunk is a significant predictor of knee injury. For upper limb, Yildirim (2010) emphasized the importance of accurate motor recruitment and precise coordination of the glenohumeral and scapulothoracic muscles in force generation and dynamic protection of the shoulder joint during functional tasks. Any change of the motor recruitment pattern may affect joint alignment and force around the glenohumeral joint and will lead to tensile overload to shoulder structures (Cools, et al., 2003; Yildirim, et al., 2010). Further complicated by the highly-repetitive nature of the lunge attack in wheelchair fencing, the risk for developing various upper limb disorders may also be compounded.

While a h igher physical demand is expected in the wheelchair fencers' upper extremities, particular in those with poor trunk control, very little is known about the motor recruitment patterns and the level of muscle activity during wheelchair fencing. EMG information on the contractile pattern, timing of activation and the relative upper limb muscle loading during wheelchair fencing was little known; all are important information to explore the motor characteristics and the possible causes of upper limb injuries in wheelchair fencers.

2.8 Wheelchair fencers at risk

A considerable number of studies had examined the complex coordination of the trunk and upper extremities during reaching in able-bodied subjects (Dean, et al., 1999; Kaminski, et al., 1995; Kaminski, 2007). Findings generally agreed that reaching movement in seated position involved a tight coupling between trunk and arm (Dean, et al., 1999; Kaminski, et al., 1995; Kaminski, 2007). The fencing lunge attack motion is a fast reaching and pointing task that requires a lot of trunk and upper limb coordination. Kaminski, et al. (1995) studied fast pointing movements to five target locations, two of which were within arm's length while three were beyond arm's length. When a target was placed beyond arm's length, trunk movement was significantly involved. Electromyography study by Dean, et al. (1999) demonstrated that lower limb muscles also actively contributed to support the body mass when self-paced reaching task was performed at the long reaching distance. The coupling of trunk-arm and lower limb reflects the multilinked structure of the body during the reaching task. The body work together to maintain a good sitting stability and a balanced body mass over the base of support in order to allow the hand to move to the desired target.

Motion analysis of the upper limb reaching task in sitting was also performed and compared in hemiparetic and healthy subjects (Hsu, et al., 2005). Subjects were asked to reach a target that was placed within and beyond the length of arm and their functional reach abilities and the ankle electromyographic data were measured. The ankle electromyographic activity patterns in hemiparetic subjects showed the muscles in the affected ankle could not be recruited timely and efficiently for the reaching task (Hsu, et al., 2005). The disintegration and reduction of the arm-trunk coupling was clearly demonstrated in the study. Such findings, although performed on hemiplegic subjects with brain lesions, the results may be reasonably applied to other diseased groups, particularly those who have lost of lower limb and trunk control such as the wheelchair fencers.

The maintenance of postural stability at seated position – an important factor in performing upper limb reaching tasks, is termed postural adjustment mechanism (Hodges & Richardson, 1997). During limb movements, activity of trunk muscles generally occurs before the arms in order to prepare the spine for perturbation (Bouisset & Zattatara, 1981; Hodges & Richardson, 1997). Electromyographic activity showed specific pattern for each focal upper limb movement (Friedli, et al., 1984). In Friedli's study (1984), the EMG of erector spinae, rectus abdominis, quadriceps femoris, hamstrings, tibialis anterior and gastrocnemius all showed their activities happened prior to the arm muscles in a specific order of activation. The onset of rectus abdominis clearly preceded the prime mover. Transversus abdominis is found to be the first trunk muscle active regardless of the direction of limb movement (Hodges & Richardson, 1997). Hodges & Richardson (1999) also found that early activation of transversus abdominis and obliquus internus abdominis occurred in both the fast and intermediate speed of the upper limb movement. For paraplegics, whose equilibrium is less stable in comparing to able-bodied subjects, they could not develop postural adjustments adapted to the perturbation; making the voluntary movement less efficient (Bouisset & Do, 2008).

Some studies have shown that the higher the vertebral lesion the subjects had, the greater decrease in the multidirectional seated postural stability (Desroches, et al., 2013; Gauthier, et al., 2013). The diminished stability of the abdominal and lower back muscles negatively affect the dynamic stability, in turn affect the strength-generating ability and the muscle synergies at the upper limbs especially those involving the thoraco-humeral muscles. Such individuals need to develop new strategies to address the loss of trunk stability while performing functional tasks. Applying these findings to Category B wheelchair fencers who have lost of lower limb and trunk control; it is reasonable to hypothesize that a greater change in motor strategy as compared to Category A, include a larger difference of upper limb movement during the lunge attack at longer fencing distance.

2.9 Methods of assessment for physical risk factors

It is apparent from the previous sections that physical loading to the upper limbs in wheelchair fencing must be higher than the able bodied fencers in view of the fact of the elimination of the footwork, and the disturbance of the kinetic chain in the whole fencing execution. Thus, it is essential to explore an objective method to document the physical loading to the upper limbs.

Identifying the proper risk factors is crucial for designing and implementing effective injury prevention intervention. Risk factors might render the athlete to be susceptible to injury; however, as suggested by Meeuwisse (1994) the mere presence of these risk factors was not sufficient to produce injury. Meeuwisse described the inciting event, as the final link in the chain that causes an injury, was usually directly associated with the onset of injury. To comprehend the possible inciting event of sport injury, Bahr & Krosshaug (2005) proposed a multifactorial model that included the fundamental understanding of the sports movement science (Figure 2.9). Detailed biomechanical description including joint motion, force and moment could provide useful information to elucidate the underlying injury mechanism. In the field of sports science, several biomechanical assessments including kinematic, kinetic and muscle activity analyzes are widely-used for motion study and documentation of the physical work load.

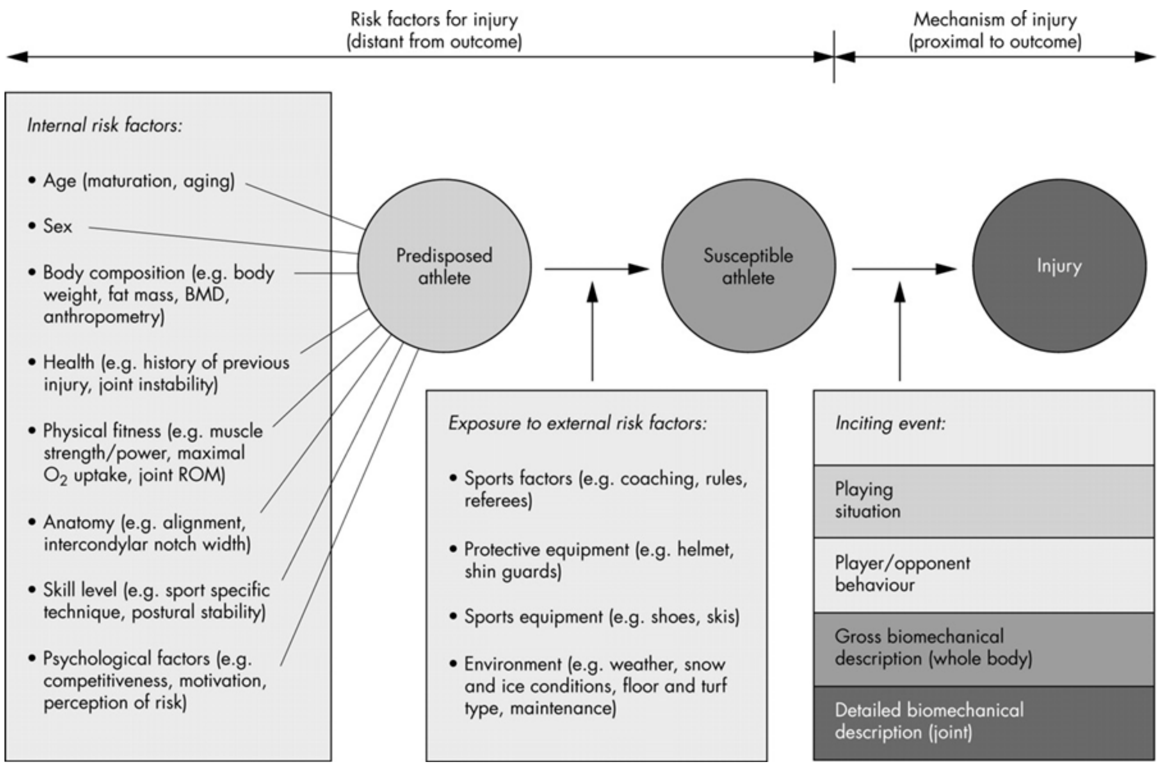


Figure 2.9 Model for injury causation (Bahr & Krosshaug, 2005)

2.9.1 Kinematic analysis for upper limb motion

Kinematic analysis is amongst the most popular used method for motion analysis in sports. Kinematic parameters describe movements of the body through space and time (e.g. position, linear and angular displacements, velocity and acceleration) regardless the causing forces. Motion analysis for upper limb, however, is relatively new compared to lower limb application. Currently, there are a few different motion analysis technologies available for upper limb biomechanical studies. They include inertial sensor, electromagnetic and optical tracking methods. Each has its advantages and disadvantages.

2.9.1.1 Inertial system – accelerator and rate gyroscope measurement

The inertial system utilizes accelerometers and rate gyros for measurement of sports movement. The technology has recently been made available commercially. Inertia system utilizes body-mounted sensors that are adhered to different body landmarks of subjects to capture data. By feeding the measured data into a series of mathematic formulas, the time-history of the position and orientation of the body parts could be reconstructed (Luinge 1999; 2005). Inertia system is able to capture and compute kinematic data include joint angles, as well as linear and angular velocity and acceleration of the body segments. Inertial systems have some specific advantages. Inertia data would not be interfered by the environment, for example, sunlight or obscuring of markers either by the body segments or by equipment. Study by Mayagoitia (2002) shown that accelerometer and rate gyroscope method was comparable to the optical method in two dimensional measurements. Since the results showed the two methods produced comparable validity results, the authors suggested that inertia method was an inexpensive and accurate alternative to the costly optical system. However, the study was done for comparing validity in two-dimensional gait analysis; the possibility to extrapolate the results to three-dimensional motion analysis is still unknown, especially if it is for upper limb motion analysis. Some studies have recently been conducted on upper limbs motion analysis (Thesis, et al., 2007; Zhou, et al., 2006), but the results on validity and reliability for sports movements that involve high speed actions are yet to be seen. Besides, the size and weight of the inertial sensors may hinder normal joint movements, especially to smaller distal joints of the upper limb. Although miniature sensors are started to be available in market, this technology for sport motion analysis is still in its infancy.

2.9.1.2 Magnetic system – electromagnetic tracking system

Electromagnetic tracking has also been used to monitor three-dimensional information of spine (Lee, 2001). Electromagnetic tracking utilized sensors which attached to the body parts of subjects and a source that generates low frequency magnetic field. The three dimensional coordinates of the sensors are detected and computed using the relative intensity of magnetic field signal. As the body part moves, the sensors will detect and send the change in magnetic signals to the computer to determine the resultant coordinates of the sensors. In turn, the angular velocity and acceleration of that particular body part can be calculated. A number of studies had been conducted and documented the high reliability of the electromagnetic tracking system (Jasiewicz, et al., 2006; Luinge & Veltink, 2005; Sabatini, et al., 2005; Saber-Sheikh, et al., 2010). However, there are several disadvantages of the magnetic systems. First, most magnetic systems require the sensors to be wired; the attached cables inevitably hinder the subject's movements thus affect the performance. Despite a few models offer wireless systems, the wireless sensors are more massive; as a result, also hinder the tracking precision. Second, any metal objects that are within the capture area would interfere the magnetic field (Milne, 1996). Although correction for metallic distortion is possible, the process is time-consuming and complicated (Lee, 2003).

Electromagnetic tracking system is deemed to be inappropriate for fencing motion study for two main reasons. One, the lunge attack in fencing is a high speed explosive motion. Not only the cables of the sensors may hinder the natural performance of fencers, the fast and sudden motion would inevitably cause significant movement of cables which in turn might detach from the sensors. Two, and more importantly, the metal body of the fencing weapon is an absolute contra-indication to the magnetic system.

2.9.1.3 Optoelectrical methods

Three-dimensional (3D) video analysis is the most commonly used motion analysis technique in sports medicine. Video based optoelectronic systems are often thought of as the laboratory gold standard (Cuesta-Vargas, et al., 2010). This system utilizes retro-reflective markers visualized by multiple high-speed video cameras; allowing the visualization of multiple body regions. Optical system has been commonly used in gait analysis with its merit of highly reliable and capable to generate three dimensional kinematic outcomes (Sabick, et al., 2005). With the success of 3D gait analysis, there is a surging demand for a standardized and reliable protocol to objectively measure the upper limb motion in clinical population. However, upper limb motion analysis has inherent problems due to the large degrees of freedom, complex multi-joint structure and lack of unconstrained cyclical movement task of the upper limbs (Mackey, et al., 2005; Rau, et al., 2000). Other factors such as the inertial effect and oscillation of external markers over the underlying skeletal or soft tissue structures, especially during rapid and jerky motions, may further impede the accuracy of upper limb kinematic measurement (Rab, et al., 2002; Reid, et al., 2010). Nonetheless, with the advancement of modern technology, development of various kinematic models (An, et al., 1991; Rab, et al., 2002) and standardization of the motion analysis methods (Wu & Cavanagh, 1995), many of the aforementioned technical difficulties were moderated. To date, more researchers start to adopt the 3D kinematic motion technology to investigate simple upper limb movement (Mackey, et al., 2005; Rau, et al., 2000). Clinical studies were conducted to examine the validity (Lempereur, et al., 2012; Rab, et al., 2002; Subramanian, et al., 2010) and reliability (Reid, et al., 2010; Mackey, et al., 2005) of the optical method for different upper limb functional activities. Rau, et al. (2000) appraised the repeatability of the optical motion analysis method for patients with brachial plexus palsy. Mackey, et al. (2005) also evaluated the 3D kinematic analysis of shoulder and elbow joints during functional tasks for children with hemiplegia and rated the repeatability to be moderate to high.

In sports medicine, high-speed video capturing system is the most commonly used method to capture kinematic data. The biggest advantage of optical system in biomechanical analysis in most sports motion is the feasibility of using large number of optical reflective markers simultaneously over various body segments, thus capturing complex body movements with high accuracy. Also, the sampling frequency for optical method could be high to support fast-speed motion. The optical tracking system is a superior choice in measuring an intricate fencing movement which involves the entire shoulder complex and upper limb.

Although the repeatability of optical motion analysis is high, the application has some limitations. First, the complete set of equipment is expensive. Second, the whole application can only be used within a confined area due to the setup of camera and video. Usually the setup has to be in a laboratory setting. Third, the placement of markers can be challenging. If the marker is obscured from the video camera, errors can be introduced and the kinematic analysis would not be accurate. Researchers (Chiari, et al., 2005; Della Croce, et al., 2005; Kadaba 1989; 1990; Leardini, et al., 2005) also indicated that errors in markers reapplication could be the major source of inaccuracy in gait analysis. While errors may exit, researchers recommended that the accuracy of optical system could be minimized by precise calibration, adjustment of camera for maximize marker visibility and checking the potential change of length of limb segments during testing (Anglin & Wyss, 2000).

2.9.1.4 Kinematic analysis for fencing motion

Kinematic studies on w heelchair fencing and able-bodied fencing are in paucity, available literatures on fencing motion analysis were all utilizing the optical method (Table 2.7). The only wheelchair fencing kinematic study by Fung, et al. (2013) had utilized the 2-dimensional video method to measure the trunk angles and speeds during lunge attack. The video motion method could be considered to be more portable that could be used outside laboratory setting. Without applying any markers or wire to the athletes, video method causes less interfere and is suitable for motion analysis during competition. However, this method could only measure and derive kinematic variables in one single anatomical plane. For detail motion assessment such as the highly mobile shoulder joint, the application of 2D video motion method would be rather limited. Also, the subjective marking of body landmarks for motion computation is always argued to lower its reliability (Anglin & Wyss, 2014). Obstruction of landmark by opponent or sports equipment may even limit its application in sports.

As discussed in the preceding section, the application of 3D optical method for wheelchair fencing appears to be a feasible method to assess the physical risk factors. However, its application to upper limb motion analysis is relatively new. There is no reliability and validity data for fast and briskly movement of wheelchair fencing. It is essential to quantify and establish the reliability and validity of optical method for wheelchair fencing upper limb motion before application.

Table 2.7 Kinematic analyses in able-bodied and wheelchair fencing

Study	**Method**	**Action**	**Captured Parameter**
Sapega, et al., 1977	Cybex II isokinetic dynamometry system	Not specified	Strength, endurance, power output
Harmenberg, et al., 1991	Goniometric switch	Lunge action	Reaction time, movement time
Cronin, et al., 2003	Linear transducer	Lunge	Horizontal displacement
Stewart & Kopetka, 2005	Peak motus analysis	Fencing	Lunge speed, maximum angular velocity
Lopez, et al., 2007	Video camera	Lunge action	Speed of weapon
Frere, et al., 2008	Vicon system EMG	Flēche attack	Time of maximal muscle activity
Gholipour, et al., 2008	High speed camera Stereo-photogrammetry	Fencing Lunge	Joint angles, stride length, travelling distance, speed of the lunge motion
Nuesch, et al., 2008	Vicon MX, 240 Hz EMG Force plate	Fleche lunge	Ground reaction force Muscle activity Kinematics of the whole body
Mantovani, et al., 2010	Vicon system	Fencing motion pattern	Acquisition
Frere, et al., 2011	Vicon EMG	Fleche attack	Velocity, angular Velocity
Morris, et al., 2011	Vicon Mx System Force platform	Fleche attack	Angular velocity, moments of force, power
Aquili, et al., 2013	Two high-speed digital cameras	Match Analysis	Different kinds of actions, number of direction change, number of lunge
Fung, et al., 2013	Peak Motus Motion Measurement System	Lunge Fast-return	Trunk velocity, maximum trunk angle during wheelchair fencing
Gutiérrez-Dávila, et al., 2013	Vicon-460 infrared camera Force platforms	Lung attack	Reaction time, force

2.9.1.5 Kinematic variables

2.9.1.5.1 Temporal kinematic data

Motion analysis studies generate kinematic data that could determine the position, angular displacement, velocities and accelerations of a body segment. Angle-time curves are often used to indicate the sequence and interplay of angles throughout a task. By referring to the various kinematic data, upper limb motion during wheelchair fencing could be quantified.

2.9.1.5.2 Cross-correlation

As stated in Section 2.7.2.2, the concept of kinetic chain may provide some hints that related the effect of truncated lower limb and trunk to the altered kinematic in wheelchair fencers. Unfortunately, quantitative analysis of kinetic chain is always limited to sophisticated biomechanical modeling that limits its use for clinical research. Also, a major limitation of previous research was that only the relative magnitudes of the body segments during the motion analysis were examined. The similarity or dissimilarity of the temporal patterns was rarely investigated. The information on t he phase relationship between the movement-time curves of the two joints is lacking yet crucial to bring insights into coordination strategies of the body segments. As such, the current study intended to utilize the mathematical technique of cross-correlation to investigate the above kinematic information (Li & Caldwell, 1999; Li & Wong, 2002). As one of the well-established approaches for comparing signals, cross-correlation is widely used in various fields including audio-signal processing and image processing. For the EMG application, cross-correlation has been used to examine myoelectric cross talk (Lowery, et al., 2003), synchronization of motor unit firing (Loeb, et al., 1987), mechanical positions (Lee & Wong, 2002) and EMG amplitudes (Li & Caldwell, 1999).

2.9.2 Motion analysis by electromyography

In sport science, it is important to establish the basic knowledge on the temporal muscle activity in different sport-specific tasks for injury prevention, training and rehabilitation. Muscle functions could be tested by different methods depending on the complexity. To understand the function of a muscle during a simple single joint movement, a dynamometer can be used to measure the muscle contraction strength or torque. However, for a more dynamic task that involves a multiple-joint movement, more sophisticated method is needed. Electromyography (EMG) is a method that records the neuromuscular activity during the entire movement pattern; including information on inter-muscular coordination and the muscles firing pattern for motion analysis. This information is essential for understanding complex muscle functions, especially in the area of disabled sports.

2.9.2.1 EMG analysis for fencing motion

EMG analysis in able-bodied fencing is emerging. However, the application had mainly focused on motor recruitment sequence, with majority over the lower limbs (Table 2.8). In wheelchair fencing, there is no study conducted to investigate the upper limb motor characteristics and activity level. EMG analysis for upper extremities during wheelchair fencing is warranted to improve the understanding of the injury mechanism.

Table 2.8 EMG studies in fencing

Study	Method	Action (weapon)	Investigated muscles
William & Walmsley, 2000	Surface EMG	Lunge attack (foil)	Anterior deltoid, triceps, bilateral biceps femoris, bilateral rectus femoris
Frere, et al., 2008	Surface EMG	Flēche attack (epee)	Deltoideus anterior, infraspinatus, triceps brachii
Nuesch, et al., 2008	Surface EMG	Fleche attack (not specified)	Tibialis anterior, gastrocnemius medialis, vastus medialis, rectus femoris, semitendinosis
Suchanowsbi, et al., 2011	Surface EMG	Lunge attack (foil)	Extensor carpi radialis Rectus femoris (rear and front legs)
Frere, et al., 2011	Surface EMG	Fleche lunge (epee)	Deltoid pars claviculars, infraspinatus, triceps brachii
Guilhem, et al., 2014	Surface EMG	Fleche attack (epee)	Soleus, gastrocnemius lateralis, tibialis anterior, vastus lateralis, rectus femoris, semitendinosis, biceps femoris, gluteus maximus

2.9.2.2 Theories and principles of EMG analysis

EMG is a biomedical signal that measures electrical currents generated in muscles during its contraction. The signal can be captured by using surface or indwelling tramuscular electrodes. Surface EMG (SEMG) applies non-invasive electrodes to the skin and captures muscle activity data, while intramuscular EMG applies needle or wire electrodes through the skin to the muscle fibers. Although intramuscular electrode can provide more accurate data, especially for deep muscles, it is generally not desirable for studying muscle coordination as the technique requires lacerating muscle tissue, causing pain upon movement and thus limiting the number of muscles being recruited (Frigo & Shiavi, 2004). SEMG on the other hand provides easy access to the physiological process that happens in superficial muscle (De Luca, 1997).

Quantifying electrical activity from muscles is not easy. Signal artifacts can occur within EMG instrumentation and during data capture. Crosstalk is the contamination of EMG signal by nearby muscle's electrical activity and is one of the most important sources of error in interpreting SEMG (Hug, 2011). To minimize crosstalk, proper location of the surface electrodes in the center of the muscle belly is required (Hermens, et al., 2000). Another artifact involves the background signals from the surrounding the electronic equipment within the testing environment. To maximize the quality of EMG signal, the signal-to-noise ratio and distortion of signal need to be minimized (Reaz, et al., 2006). The reliability of SEMG on dynamic movements like lunge attack in fencing could also be affected by many other factors, including the motion artifact from electrode contact, cabling and skin impedance. Although thorough preparation combined with clear and standardized protocol can help minimizing the risks, the reliability of EMG measurement specifically for wheelchair fencing has not been established. Part of the current research study is to establish the repeatability of the SEMG protocol prior to our testing to ensure the SEMG is a suitable and reliable tool for our research.

2.9.2.3 Normalization procedures

To evaluate and compare EMG data obtained from different subjects or from the same subject on different days, a normalization procedure is usually necessary for both recording and quantifying the EMG data. Normalization is the process of establishing a reference point of EMG for comparison. Data would be referenced to some standard values for different muscles to enable comparison between different individuals. There are two parts involved in normalization: amplitude normalization and time normalization. Each serves a different purpose.

2.9.2.3.1 Amplitude normalization

There are several methods for EMG amplitude normalization. The most common and valid method is the maximal voluntary isometric contraction (MVIC). The EMG is recorded from a muscle during a MVIC. The data will be used as a reference point for that individual muscle. All other EMG measures from that muscle will be compared against the reference data point. To obtain MVIC EMG, subject will perform maximum muscle contractions for at least 3 times, an interval of at least 2 minutes between repetitions is given to reduce any fatigue effects (Mathiassen, et al., 1995). The highest value obtained from all repetitions is the reference value for normalizing EMG signals. All EMG signals are rescaled into a percentage of MVIC to enable direct quantitative comparisons. The reliability of MVICs within individuals on the same day is reported high (Bolgla & Uhl, 2007; Dankaerts, et al., 2004). High repeatability requires proper guidance of the subjects to perform the tests identically with each repetition, familiarity of the subjects with the production of maximum effort and the avoidance of fatigue. In the present investigation, in order to ensure MVIC was not under reported, all wheelchair fencers included in the study were able to lunge pain-free fencing attacks. Proper guidance were provided to ensure subjects' familiarity to the test and ample of rest time between tests were given to avoid fatigue.

2.9.2.3.2 Time normalization

For better understanding of the movement pattern, comparing the sequence of muscle firing is important. In order to obtain a representative EMG profile, consecutive movement cycles (or trails) are usually averaged. However, due to the difference of cycle duration in each movement cycle, it is first necessary to interpolate the cycle to achieve an equal number of points for subsequent averaging (Hug, 2010). In general, 100-400 points are commonly used (Shiavi & Green, 1983). In the present investigation, each lunge attack trail was time-normalized into 100 equally portioned segments. The onset and the occurrence of peak EMG as expressed in percentage of lunge cycle were computed to compare the motor pattern for both Category A and B fencers.

Once EMG time and amplitude normalization is completed, the EMG parameters could be measured and compared.

2.9.2.4 EMG parameters

Most of the SEMG are used for measuring the timing of muscle activation, amplitude and frequency domains. The application of these variables has been justified by a huge literature for both static and dynamic muscular works (Giroux & Lamontague, 1990; Zijdewind, et al., 1995).

2.9.2.4.1 Timing of activation

There are two sets of EMG data related to timing of activation are of interest - onset time and time to peak EMG. The onset time of an EMG signal refers to the time to detect an EMG signal when a muscle starts contraction from a complete relaxed position. This is usually interpreted as the reaction time of the movement. The time to peak EMG refers to the duration from the beginning to the peak EMG amplitude value during a muscle activity; the peak EMG time illustrates the firing pattern for the maximal muscle activity.

2.9.2.4.2 Amplitude domain

Within the amplitude domain, two different types of information are available. The strength of muscle contraction and the total muscle activity. The amplitude of EMG is related to the force or strength that a muscle may generate. However, absolute peak EMG value is rarely used as a measure of strength; as peak EMG is only meaningful when a muscle contraction produces an averaged curve. A raw EMG signal rarely produces a single peak unless there is an artifact. Instead of the peak EMG, the average peak EMG is usually used for a more reasonable interpretation. In the present study, the peak EMG in the 5 trials was averaged for comparison. The other amplitude related measure is the integrated EMG (iEMG), which is the integral area under the EMG amplitude for a defined period of time. It directly represents the total muscle activity for the movement during that period of time.

2.9.2.4.3 Frequency domain

When EMG signal is presented as function of time, the change of the profile of muscle activation over time can be obtained. Any shift in the median frequency of EMG signal is an indication of muscle fatigue (Basmanjuan & De Luca, 1985; De Luca, 1997). Given the major objectives of this study is to quantify the motor sequence and compare the muscle effort between the Category A and B fencers, the objective of the current research did not include studying muscle endurance or effect of muscle fatigue on fencing performance, the frequency EMG domain was hence not included.

2.10 Rationale of the present study

The health benefits of sport activities for individuals with disability are numerous and apparent. As wheelchair fencing gains popularity in the sports arena, the associated risk of injury while participating in sports, especially for athletes competing at elite level also increases. The high injury prevalence reported indicated that the current intervention strategies in the prevention or rehabilitation of the injuries in elite disable athletes are not effective. Many injury surveillance studies were conducted, however, different definitions, durations, and methodologies were used, making comparison challenging. In able-bodied fencing, fencers execute the lunge attack by highly-coordinated and powerful lower leg muscular motions. With the elimination of footwork, wheelchair fencers present a very different movement pattern. Some studies showed that an elimination of footwork would result in substantial reduction of upper limb joint speed, increase in joint angles, higher in muscle activity level and alteration of motor patterns. Such phenomenon is colossal in individuals with lower trunk function. Thus, the elimination of the footwork to perform the fencing tasks would possibly increase the physical risk factors to the upper limbs component. Study on the identification of the physical risk factors associated with wheelchair fencing is rare. Many kinematic and EMG studies conducted either on able bodied fencers or to other wheelchair athletes provided preliminary information on movement patterns, however, the results are not directly applicable to wheelchair fencers. Although it is apparent from literature reviews that physical loading to the upper limbs in wheelchair fencing must be higher than the able bodied fencers, an objective method to document the loading has yet to be established. Optical tracking method and EMG were commonly used in sport medicine studies, however, it was never used in assessing the fast, sudden upper limb fencing actions; their reliability and validity for the fast and briskly movement of wheelchair fencing are unknown. Our literature review demonstrates the gap in research areas in the documentation of the injury profile in wheelchair fencers, identification of wheelchair fencing injury risk factors, and the establishment the reliability and validity of the optical tracking and EMG methods for assessing the physical risk factors. This thesis aims to address such gaps.

2.11 Objectives of the present study

The main objectives of the present investigation were to: 1) examine the injury pattern of wheelchair fencers, 2) quantify the physical risk factors using kinematic and electromyographic approaches, and 3) identify the physical risk factors of the Category A and Category B wheelchair fencers.

2.12 Specific aims of different research experiments involved in the current thesis

The specific aims of each research studies are to:

1. Identify the injury incidence, pattern and severity of the common musculoskeletal injuries that affects the elite Chinese male foil wheelchair fencers;
2. Establish the feasibility of using optical motion and EMG analysis methods to document the lunge attack motion;
3. Quantify the physical risk factors during lunge attack at different fencing distances by kinematic and electromyographic approaches;
4. Compare the characteristics of the physical risk factors between Category A and Category B fencers

2.13 Hypotheses

This doctoral project was designed to test the following four hypotheses:

1. Injury pattern was different between able-bodied fencers and wheelchair fencers
2. Injury incidence and pattern was higher in Category B fencers than Category A fencers
3. Kinematic parameters were different between Category A and Category B fencers
4. Motor recruitment pattern and EMG activities were different between Category A and Category B fencers

2.14 Clinical relevance of the present study

The results of current research project will lay a foundation for better understanding the biomechanics of a common wheelchair fencing technique (a lunge attack). The findings may provide insights to the development of relevant injury prevention program, training regime and rehabilitation strategy for wheelchair fencers. The results would also enhance the development of sports science and medicine for other wheelchair sport events.

CHAPTER 3

Musculoskeletal injuries in elite able-bodied and wheelchair foil fencers

3.1 Introduction

Able-bodied fencers are vulnerable to a wide range of musculoskeletal injuries of their lower extremities because of the repetitive, asymmetrical and impulsive nature of the sport (Moyer & Konin, 1992; Roli & Bianchedi, 2008). Since wheelchair fencers compete in a fastened wheelchair, they solely rely on their arms and trunks for perform all necessary actions; their upper body loading is likely to increase, particularly among those with poor trunk control. Epidemiological study on wheelchair fencing related injuries, however, have been scanty. Following the suggestion by van Mechelen, et al. (1992) in the sports injury prevention, determining the prevalence and severities of the injuries associated with that sport is the fundamental step. Accordingly, this study was conducted to (1) identify and compare the injury patterns of the Hong Kong elite able-bodied and wheelchair foil fencers and (2) compare the injury pattern in wheelchair foil fencers with good trunk control (Category A) and without active trunk control (Category B).

3.2 Methods

3.2.1 Participants

Eighteen wheelchair fencers and 12 able-bodied foil fencers were recruited from the Hong Kong able-bodied and the Paralympic squad respectively. However, four wheelchair fencers and two able-bodied fencers withdrew from the study at 6 months to one year after the commencement of the study due to personal reasons. A total of 14 wheelchair fencers and 10 able-bodied elite foil fencers completed the 3-year prospective injury surveillance study.

All recruited fencers must be at elite level who received the elite training scholarship scheme as provided by the Hong Kong Sports Institute. Wheelchair fencers without permanent International Wheelchair Fencing Committee (IWFC, 2010) classifications were excluded. Written consent was obtained for each participant and ethics approval for this study was granted by the Ethics Committee of the Hong Kong Polytechnic University (Appendix I and II). Demographic characteristics of the participants are shown in Table 3.1.

Table 3.1 Demographic of the study sample (mean ± standard deviation).

	Wheelchair fencers (n=14)	Able-bodied fencers (n=10)
Age (years)	28.6 ± 6.8	27.0 ± 5.5
Gender (Male: Female)	7: 7	10: 0
Fencing experience (years)	10.1 ± 5.3	10.2 ± 3.8
Body mass index (kg/m^2)	19.8 ± 3.5	18.6 ± 3.1
Classification status	Category A: 7; Category B: 7	Not applicable
Ambulatory level	Walker: 7; Wheelchair user: 7	Not applicable

3.2.2 Study design and measurements

The 3-year prospective investigation on the incidence of injuries among Hong Kong elite able-bodied and wheelchair fencers was conducted between November 2006 and October 2009. All fencers were first interviewed by three trained physiotherapists independently to establish the baseline information included age, fencing experience, category of disability and history of previous injuries. The fencers were interviewed monthly by physiotherapists to collect data regarding training duration, match duration and injury characteristics including location, type, severity, nature (newly-acquired or recurrent) and cause (trauma or overuse) of injury. A standardized injury registration form was created for such purpose (Appendix III and IV). All reported injuries were examined by the chief investigator (a registered physiotherapist) and an orthopedic specialist from the Hong Kong Sports Institute Sports Medicine Center to provide diagnoses. A mandatory four-hour orientation and training session was provided to all the participating investigators in order to standardize the interview and data collection procedures. The training included discussion of definitions and interview skills. For quality control, the chief investigator reviewed all interview results for the first three months. This review process was continued until the chief investigator found no evidence of inter-tester discrepancy in subsequent interviews.

3.2.3 Definition of injury

In this study, an injury was defined as a damage or trauma which occurred during training or competition that caused at least 1 day absence from training or competition. Incidence rate of injury was calculated as the number of injuries per 1000 hours of exposure. The athletic exposure was defined as the duration of practice or competition in fencing. In order to ensure our findings are comparable to other studies, we adopted the definition of severity of injury from the Athletes with Disabilities Injury Registry (Ferrara & Buckley, 1996). The severity of injury is categorized by the number of days the athlete could not participate in the sport due to injury. It was classified as minor if there was a loss of 7 or less participation days, moderate for 8–21 loss days, and major if 22 and more loss days. Detail definitions of injury characteristics are shown in Table 3.2.

Table 3.2 Definitions of injury characteristics

Injury Characteristics	Definition
Nature of injury	
Newly-acquired	Injury occurring for the first time in the past three months
Recurrent	Injury of the same type and location, which occurred after an athlete's return to full participation from the previous injury
Causation of injury	
Traumatic	Injury resulting from a sudden incident during fencing
Overuse	Painful syndrome that has an insidious onset during fencing practices or competitions without any known history of trauma
Types of injury	
Sprain	Injury of ligaments or joint capsule without rupture
Strain	Injury of muscles or tendon tear including acute or chronic tendinopathy
Contusion	Tissue bruise caused by direct collision without a break in the skin and a subcutaneous hemorrhage
Fracture	Bony damages that resulted in discontinuity of trabeculae with radiological confirmation
Subluxation / dislocation	Partial and complete displacement of the bony parts of a joint
Rupture of ligament / tendon	Major ligamentous or tendon rupture that confirmed by ultrasonograph, magnetic resonance imaging or computed tomography scan

3.2.4 Statistical analysis

Independent t-tests were used to analyze the quantitative data between able-bodied and wheelchair fencers as well as between Category A and Category B fencers. Incidence rates among groups (able-bodied and wheelchair fencers) and Categories (Category A and Category B wheelchair fencers) were analyzed at a significance level (α) of <0.05, using MedCalc 11.3.5 (MedCalc Software bvba, Mariakerke, Belgium). The between-groups (able-bodied versus wheelchair fencers) and within-group (Category A versus Category B fencers) relative risk (RR) of injury and corresponding 95% confidence intervals (CI) in relation to nature, injured location and severity of injury were computed. Relative risk is considered significant if the 95% CI does not include the value of 1. All estimates were based on Poisson law for small number to calculate the exact p values and the corresponding 95% CIs (Woodward, 2005).

3.3 Results

3.3.1 Characteristics of subjects

No significant difference in age, fencing experience and body mass index was noted between able-bodied and disabled groups (Table 3.1). Seven wheelchair fencers belong to Category A and seven to Category B. No subject in this study was classified as Category C. Medical conditions of wheelchair fencers included lower limb amputation (n=2), spinal cord injuries (n=6), cerebral palsy (n=2), poliomyelitis (n=3) and congenital limb deficiencies (n=1).

3.3.2 Incidence of injury

During the 3-year study period, all fencers had at least one injury that caused them absence from training or competition. Wheelchair fencers had statistically higher injury rate than the able-bodied fencers ($p<0.01$). Totally, 95 injuries were recorded over the 3-year period, and with an incidence rate of 3.85 injuries/1000 hours, 95% CI: 3.12-4.71 among wheelchair fencers versus 62 i njuries in able-bodied fencers with 2.41 injuries/1,000 hours, 95% CI: 1.85-3.09 (Table 3.3). Both fencer groups sustained significantly more injuries during competition than in training ($p<0.05$) (Table 3.3). Category B wheelchair fencers were more susceptible to injuries (4.87 injuries/1000 hours; 95% CI: 3.67-6.34) than Category A fencers (2.99 injuries/1000 hours; 95% CI: 2.14-4.08), $p< 0.05$ (Table 3.4).

Table 3.3. Total exposure time (hours), number of injuries and injury incidence (95% CI) per 1000 player hours of able-bodied and wheelchair fencers

	Able-bodied (n=10)	Wheelchair (n=14)
Training		
Exposure	22147	22276
Injuries	44	83
Incidence	1.99 (1.44, 2.67)	3.73 (2.97, 4.62)*
Competition		
Exposure	3552	2388
Injuries	18	12
Incidence	5.07 (3.00, 8.01)†	5.03 (2.60, 8.78)†
Total		
Exposure	25699	24664
Injuries	62	95
Incidence	2.41 (1.85, 3.09)	3.85 (3.09, 4.71)*

†within-group difference between training and competition at p<0.05 significance level
*between-groups difference at p<0.05 significance level

Table 3.4 Total exposure time (hours), number of injuries and injury incidence (95% CI) per 1000 player hours of Category A and Category B wheelchair fencers.

	Exposure	Injuries	Incidence
Category A fencers	13368	40	2.99 (2.14, 4.08)
Category B fencers	11296	55	4.87 (3.67, 6.34)*

*between group difference at p<0.05 significance level

3.3.3 Causation of injury - newly-acquired versus recurrent

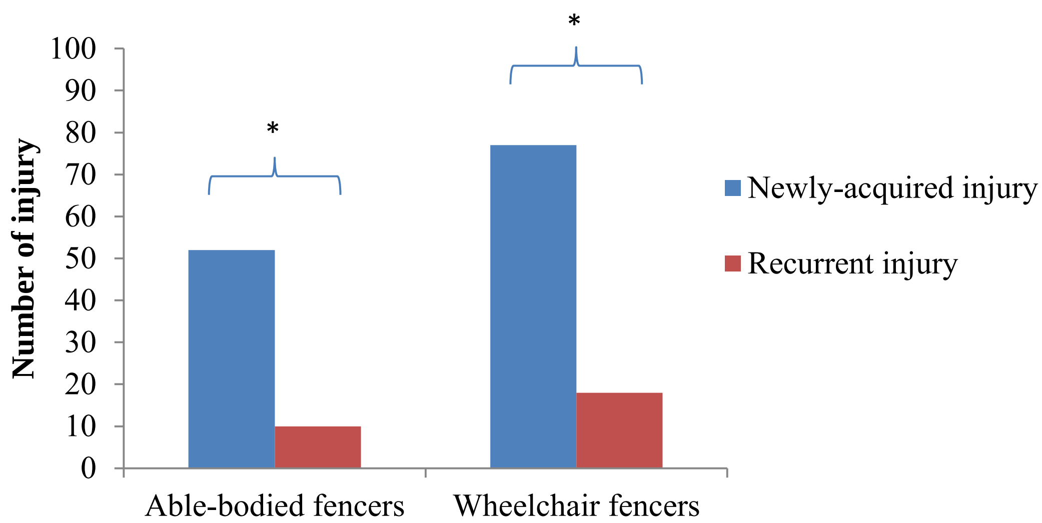
In the wheelchair fencers group, 77 out of 95 injuries (84.8%) were newly-acquired and in the able-bodied fencers group, 52 out of 62 injuries (81.0%) were newly-acquired (Figure 3.1). Within group analysis showed that newly-acquired injury was more frequent than recurrent injury, $p<0.01$. In wheelchair fencer group, the rate of newly-acquired injuries was 3.12/1000 hours (CI: 2.46-3.90), and the rate of recurrent injuries was 0.73/1000 hours (CI: 0.43-1.15). In able-bodied fencer group, the rate of newly-acquired injuries was 2.02/1000 hours, (CI: 1.51-2.65) compared to that of recurrent injuries of 0.39/1000 hours, (CI: 0.19-0.72). There is no significant between-group difference between the able-bodied and disabled group for newly-acquired ($p=0.075$) and recurrent injury ($p=0.105$).

Amongst the wheelchair fencers, the reported injuries in Category A and B fencers were mainly newly-acquired (75% in Category A and 80% in Category B). Statistically, both fencing groups displayed a significant higher incidence of newly-acquired injury (Category A fencers: 2.24 injuries/1000 hours, 95% CI: 1.51-3.20; Category B fencers: 3.90 injuries/1000 hours, 95% CI: 2.83-5.23) when compared to recurrent injury (Category A fencers: 0.75 injuries/1000 hours, 95% CI: 0.36-1.38; Category B fencers: 0.97 injuries/1000 hours, 95% CI: 0.49-1.74), ($p<0.01$). There was no s ignificant difference in the incidence of newly-acquired ($p=0.058$) and recurrent injury ($p=0.545$) between the Category A and B fencers. 71% of all wheelchair fencers and 80% of all able-bodied fencers sustained recurrent injuries in forms of muscle strains or joint sprains. All the reported recurrent injuries were minor to moderate in severity.

3.3.4 Nature of injury - traumatic versus overuse

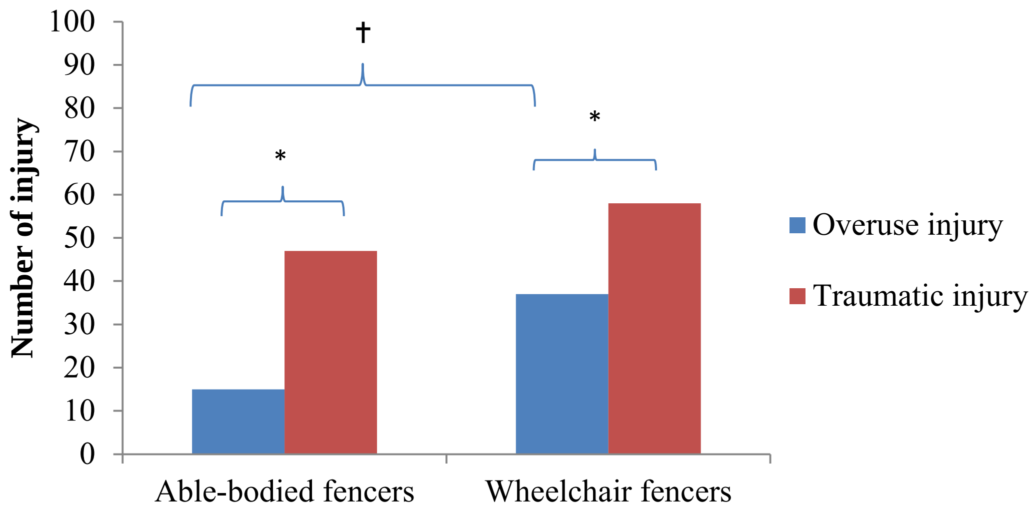
When examined the nature of injury, 58 out of the 95 injuries (75.8%) in wheelchair fencers and 47 out of 62 injuries (61.1%) in able-bodied fencers were caused by trauma (Figure 3.2). The incidence of injury caused by trauma (wheelchair fencers: 2.4 injuries/1000 hours, 95% CI: 1.79-3.04; able-bodied fencers: 1.8 injuries/1000 hours, 95% CI: 1.34-2.43) was significantly higher than that of the overuse (wheelchair fencers: 1.5 injuries/1000 hours, 95% CI: 1.06-2.07; able-bodied fencers: 0.6 injuries/1000 hours, 95% CI: 0.33-0.96) in both able-bodied and wheelchair fencers ($p<0.05$). Significant group-difference was found in the number of injury incidence that caused by overuse injury ($p<0.01$) but not for the traumatic injury ($p=0.199$).

For the two wheelchair-fencing groups, more traumatic injuries were recorded (Category A: 23 out of 40 injuries or 57.8% of the overall reported injuries; Category B: 35 out of 55 injuries or 64% of the overall reported injuries). Statistically, Category B fencers showed significant higher number of traumatic injuries (3.10 injuries/1000 hours, 95% CI: 2.16-4.31) than overuse injuries (1.77 injuries/1000 hours, 95% CI: 1.08-2.73; $p=0.043$). Whereas, there was no significant difference for the incidence of trauma injury (1.72 injuries/1000 hours, 95% CI: 1.09-2.58) and overuse injury (1.27 injuries/1000 hours, 95% CI: 0.07-0.20) in Category A fencers ($p=0.343$). The between-group analysis revealed that Category B fencers had higher incidence of traumatic injury than the Category A fencers, $p=0.026$. However, there was no significant difference in the incidence of overuse injury between the Category A and B fencers ($p=0.314$).



*within-group difference, p<0.01

Figure 3.1 Number of newly-acquired and recurrent injury in able-bodied and wheelchair fencers



*within-group difference, p<0.05; †between-group difference, p<0.01

Figure 3.2 Number of injuries in able-bodied and wheelchair fencers due to overuse injury or trauma

3.3.5 Injury location, types and severity of injury

In general, wheelchair fencers suffered higher incidence of upper extremities injuries (73.8%, 2.83 injuries/1000 hours) than able bodied fencers (16.1%, 0.39 injuries/1000 hours) (Table 3.5 and 3.6). The most common diagnoses among wheelchair fencers were elbow strain (32.6%) and shoulder strain (15.8%) (Figure 3.3 and Table 3.5). Shoulder injuries usually led to longer absence days from training/competition especially in the Category B wheelchair fencers. Four out of the 7 Category B wheelchair fencers were absent from training/competition for more than 28 days due to partial-thickness tendon tears of the shoulder on their fencing arms. Unlike wheelchair fencers, able-bodied fencers were more susceptible to sustain lower extremity injuries (69.4%, 1.67 injuries per 1000 hours), included muscle strain at knee and thigh (22.6%), ankle sprain (14.5%) and knee sprain (11.3%) (Figure 3.4 and Table 3.6). Able-bodied fencers have relatively low incidence of injury to upper limb (16.1%) and spine (14.5%) (Table 3.6). They also have more major to severe injuries of hamstring tears, anterior cruciate ligament rupture and fractured ankles; resulted in prolonged absent from training/competition (Table 3.6 and Figure 3.4).

3.3.6 Injury risk

Wheelchair fencers had higher relative risk to minor injury (RR: 2.35; 95% CI: 1.56-3.61), muscle strain (RR: 2.16; 95% CI: 1.34-3.56), shoulder injury (RR: 13.55; 95% CI: 3.39-17.76) and elbow injury (5.90; 95% CI: 2.45-17.21) than the able-bodied fencers (Table 3.7). The Category B fencers had higher relative risk of sustaining muscle strain (RR: 1.83; 95% CI: 1.04-3.28) and shoulder injury (RR: 4.97; 95% CI: 1.82-16.87) as compared to the Category A fencers (Table 3.8).

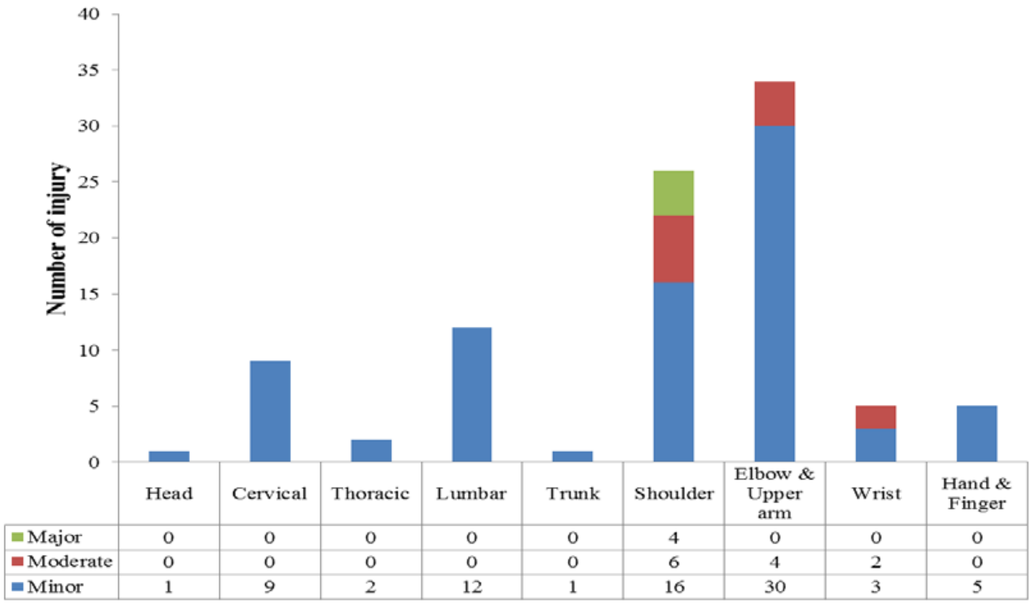


Figure 3.3 Severity of injury at various anatomical position in wheelchair fencers (n=14)

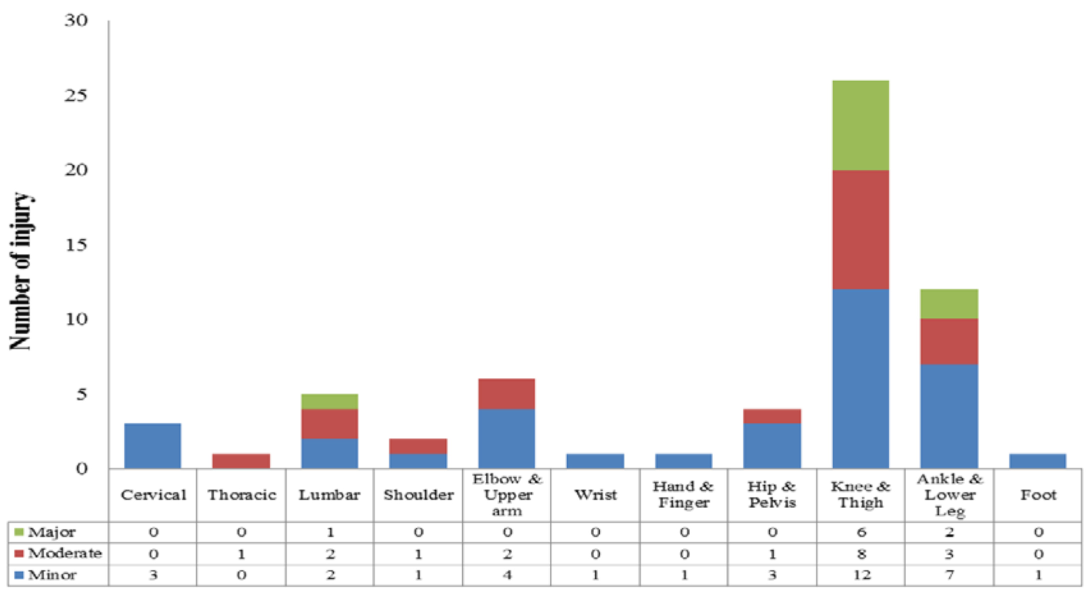


Figure 3.4 Severity of injury at various anatomical position in able-bodied fencers (n=10)

Table 3.5 Injury distribution by types and locations in wheelchair fencers (n=14)

	Location	Fracture	Dislocation	Tendon/ Ligament rupture	Meniscus/ Cartilage injury	Sprain	Strain	Contusion	Incidence per 1000 hours (Percentage)	Number of injury (Percentage)
Head/Spine	Head	0	0	0	0	0	0	1	0.1 (1.1%)	25 (26.2%)
	Cervical	0	0	0	0	5	4	0	0.4 (9.5%)	
	Thoracic	0	0	0	0	2	0	0	0.1 (2.1%)	
	Lumbar	0	0	0	0	8	3	1	0.5 (12.6%)	
	Trunk	0	0	0	0	0	0	1	0.1 (1.1%)	
Upper Limb	Shoulder andUpper arm	0	0	4	0	7	15	0	1.1 (27.4%)	70 (73.8%)
	Elbow/Forearm	0	0	0	0	3	31	0	1.4 (35.8%)	
	Wrist	0	0	0	1	2	2	0	0.2 (5.3%)	
	Hand and Finger	0	0	0	0	0	1	4	0.2 (5.3%)	
Total number of injury (Percentage)		0 (0%)	0 (0%)	4 (4.2%)	1 (1.1%)	27 (28.4%)	56 (58.9%)	7 (7.4%)	95 (100%)	

*There were no lower limb injury and the items for lower limb are omitted here.

Table 3.6 Injury distribution by type and location in able-bodied fencers (n=14)

	Location	Fracture	Dislocation/ Subluxation	Tendon/ Ligament rupture	Meniscus/ Cartilage injury	Sprain	Strain	Contusion	Incidence per 1000 hours (Percentage)	Number of injuries (Percentage)
Spine	Cervical	0	0	0	0	3	0	0	0.1 (4.8%)	9 (14.5%)
	Thoracic	0	0	0	0	1	0	0	0.1 (1.6%)	
	Lumbar	0	0	0	0	3	2	0	0.2 (8.1%)	
Upper Limb	Shoulder andUpper arm	0	0	0	0	1	1	0	0.1 (3.2%)	10 (16.1%)
	Elbow/Forearm	0	0	0	0	0	6	0	0.2 (9.7%)	
	Wrist	0	0	0	0	1	0	0	0.1 (1.6%)	
	Hand and Finger	0	0	0	0	0	0	1	0.1 (1.6%)	
Lower Limb	Hip and Pelvis	0	0	0	0	2	2	0	0.2 (6.5%)	43 (69.4%)
	Knee and Thigh	0	1	2	2	7	14	0	1.0 (41.9%)	
	Ankle and Lower Leg	1	0	0	0	9	2	0	0.5 (19.4%)	
	Foot	0	0	0	0	1	0	0	0.1 (1.6%)	
	Total (Percentage)	1 (1.6%)	1 (1.6%)	2 (3.2%)	2 (3.2%)	28 (45.2%)	27 (43.6%)	1 (1.6%)	62 (100%)	

*There were no head and trunk injury. These two items are omitted here.

Table 3.7 Injury incidence (per 1000 hours) and relative risk of injury between able-bodied and wheelchair fencers (Able-bodied fencers as the control group)

	Able-bodied fencers	**Wheelchair fencers**	**Relative risk (95% CI)**
Nature of injury			
Fracture	0.04	0	-
Dislocation / Subluxation	0.04	0	-
Tendon / Ligament rupture	0.08	0.16	2.08 (0.30-23.04)
Meniscus / Cartilage injury	0.08	0.04	0.52 (0.01-10.00)
Sprain	1.09	1.09	1.00 (0.57-1.75)
Strain	1.05	2.27	2.16 (1.34-3.56)†
Contusion	0.04	0.28	7.29 (0.94-328.72)
Body part of injury			
Head	0	0.04	-
Cervical	0.12	0.36	3.13 (0.78-17.95)
Thoracic	0.04	0.08	2.08 (0.11-122.95)
Lumbar	0.19	0.49	2.50 (0.82-9.06)
Trunk	0	0.04	-
Shoulder and Upper am	0.08	1.05	13.55 (3.39-17.76)†
Elbow and Forearm	0.23	1.38	5.90 (2.45-17.21)†
Wrist	0.01	0.20	5.21 (0.58-246.41)
Hand and Finger	0.01	0.20	5.21 (0.58-246.41)
Hip and Pelvis	0.62	0	-
Knee and Thigh	0.54	0	-
Ankle and Lower Leg	0.47	0	-
Foot	0.041	0	-
Severity of injury			
Minor	1.36	3.20	2.35 (1.56-3.61)†
Moderate	0.70	0.49	0.69 (0.31-1.52)
Major	0.35	0.16	0.22 (0.05-0.66)

† Statistically significant as determined by 95% CI

Table 3.8 Injury incidence (per 1000 hours) and relative risk of injury between Category A and Category B fencers (Category A fencers as the control group)

	Category A	Category B	Relative risk (95% CI)
Nature of injury			
Tendon / Ligament rupture	0	0.35	-
Meniscus / Cartilage injury	0	0.09	-
Sprain	1.05	1.15	1.10 (0.48-2.52)
Strain	1.65	3.01	1.83 (1.04-3.28)†
Contusion	0.30	0.27	0.89 (0.13-5.27)
Body part of injury			
Head	0	0.09	-
Cervical	0.60	0.09	0.15 (0.00-1.10)
Thoracic	0.07	0.09	1.18 (0.02-92.90)
Lumbar	0.37	0.62	1.66 (0.45-6.62)
Trunk	0	0.09	-
Shoulder and Upper arm	0.37	1.86	4.97 (1.82-16.87)†
Elbow and Forearm	1.20	1.59	1.33 (0.64-2.79)
Wrist	0.15	0.27	1.78 (0.20-21.25)
Hand and Finger	0.22	0.18	0.80 (0.07-6.89)
Severity of injury			
Minor	2.69	3.8	1.41 (0.89-2.27)
Moderate	0.30	0.7	2.37 (0.63-10.74)
Major	0	0.4	-

† Statistically significant as determined by 95% CI

3.4 Discussion

This is the first cohort study to document the injury incidence among the elite able-bodied and wheelchair fencers. Given the outstanding performance in international and Asian region tournaments, the wheelchair and able-bodied foil delegates of the Hong Kong team represent the profile of world-class elite fencers and the injury profile from our study should be able to generalize to elite fencers. Our prospective longitudinal study design with well-defined injury terminology and grading of injury severity also enabled objective comparison of the injury incidence between the two athletic groups.

3.4.1 Injuries amongst able-bodied fencers

The results of this study concurred with other previous literature that the injury incidence of able-bodied fencers is relatively low (Harmer, 2008; Naghavi, 2002; Roli & Fasci, 1998). Although the injury incidence is low, the reported range is wide; varied between 0.3 and 51.8 per 1000 exposure hours (Zemper & Dick, 2007). The discrepancy could be attributed to the differences in study design, competitive levels of recruited fencers and the terminologies used in documenting injury. Larger injury surveillance studies with prospective design are definitely warranted in future.

In concordance with most other epidemiological studies (Harmer, 2008; Naghavi, 2002; Zemper & Dick, 2007), this study revealed that traumatic injuries of the lower limbs were predominating in able-bodied fencers. Knee and thigh region (41.9%) are the most common injury sites. Muscle strain, especially in hamstrings and quadriceps, was the most common cause of prolonged absence from training or competition in able-bodied fencers (Table 3.6). The injury pattern of able-bodied fencers may be related to the biomechanics of the lunge attack. The attack begins with a powerful extension of rear leg to propel the body towards opponent then immediately followed by a deceleration to prevent excessive forward movement (Stewart & Kopetka, 2005; Szilagui, 1993). Fencers also have to retreat rapidly to avoid attack from opponent. The repetitive and impulsive nature imposes excessive loading stress to fencers' hips and knees and renders joint and muscle injuries (Bennell & Crossley, 1996). Further studies focusing on the kinetics of fencing legs will improve our understanding on their injury mechanism.

3.4.2 Injuries amongst wheelchair fencers

Shoulder and wrist injuries are common among wheelchair athletes participating in track events, road racing and wheelchair basketball. The high-intensity wheelchair maneuvering and repetitive overhead actions such as shooting and passing in wheelchair basketball inevitably create additional stress thus injury to upper extremities (Curtis & Dillion, 1985; Groah & Lanig, 2000; Klenck & Gebke, 2007). The findings of the current research on wheelchair fencers are shown to be consistent with other disabled sports studies in shoulder and elbow injuries (Table 3.5). Although there is no wheelchair maneuvering involved in the sport, wheelchair fencers still bear excessive upper arm loading given the brisk and impulsive lunge movements throughout a game. Additionally, since wheelchairs are fixated in predetermined positions, the wheelchair fencers can only approach or retrieve from opponents by using their trunks and upper limbs; together, may cause the wheelchair fencers to have higher risk to suffer from upper limb traumatic and overuse injuries.

Wheelchair fencers with poor trunk control (Category B) were shown to have higher risk to sustain the severe upper limb musculoskeletal injury, with especially affecting their shoulders, than fencers with better trunk control (Category A). This may be related to the truncated “kinetic chain”. Groppel (1992) stated that different body parts can be visualized as a system of chain links, which the force generated by one body segment will be transferred to other body parts. The activation of a kinetic chain usually initiates from lower extremities which transmit ground reaction forces and provide a good pivot system for body and arms movement. The sequential activation proceeds from legs, through the hips, trunk, scapulothoracic and glenohumeral joints, and eventually reaches the distal part of an arm (Groppel, 1992). Any deviation from the activation sequence may jeopardize the performance of the intended action and increase the risk of injury (Groppel, 1992; Kibler, 1994). Wheelchair athletes, inevitably, alter such kinetic chain due to the lack of footwork. The kinetic chain sequence is further aggravated among wheelchair fencers with compromised trunk control. The lack of leg work and insufficient pivot system was compensated by generating extra force and power through their arm muscles to create the necessary thrust, contributing to the prevailing strain injury of shoulder.

3.4.3 Clinical implications

Precisely identifying risk factors is crucial to the success of any sports injury prevention programs (Bahr, et al., 1997; Janda, 2003; van Mechelen, et al., 1992). Our study found distinct injury profiles for both able-bodied and wheelchair fencers; indicating that the injury prevention strategies from the able-bodied fencers may not directly adopt to the disabled group. Injury prevention and rehabilitation programs specific to wheelchair fencing to minimize injury and optimize recovery are needed. For able-bodied fencers, lower limb stretching and plyometric programs, as well as knees and ankles proprioceptive training are recommended to minimize strain and sprain injuries at high risk regions (Murga, 2006; Rippetoe, 2000). For wheelchair fencers, stabilization and strengthening program for shoulder and elbow muscles should be emphasized. Special attention should also be paid to Category B fencers given their increased vulnerability to severe shoulder injuries. Diagnostic imaging such as ultrasonography should be promptly ordered to rule out any rotator cuff tear if necessary.

Given the highly repetitive nature of fencing action, a high incidence of recurrent injury is expected. However, our results showed that newly-acquired injuries are more prevalent across all fencer groups. We hypothesize that this may be resulted from the premature return to sports following injury. Since most of the fencing injuries were minor to moderate in severity, some elite athletes might be motivated to return to practice or competition prematurely. Relying on uninjured parts to compensate for the deficit, the altered movement biomechanics may impose increased risk of injury to other body parts. Rehabilitation professionals should closely monitor the progress and symptoms of athletes at their initial stage of sports return.

3.4.4 Limitations of study

There are a number of limitations to this study. Although all the elite able-bodied and wheelchair foil fencers of the Hong Kong squad team were recruited, the sample size remained small. Taking the small sample size into account, we estimated the p-values and the corresponding 95% CIs from the exact values of Poisson distributions (Clarke & Cooke, 2005; Woodward, 2005). With a representable sample in this study (i.e. we had almost included the entire target population in Hong Kong prospectively), we would be able to obtain unbiased estimates.

We acknowledge there is recall bias in our study. Relying on athletes to report their injury and exposure hours on monthly basis inherited some subjectivity and errors. We could recommend implementing a training diary or web-based charting system in the future to minimize the bias.

Similar to other injury surveillance, it is difficult to eliminate confounding factors. Activities of daily living might unavoidably aggravate our athletes' symptoms and hindered their recovery as they could not have sufficient post-injury rest (Curtis, et al., 1999). Future biomechanical studies should quantify the altered kinematics and kinetics of wheelchair fencing in order to explore the potential injury mechanism.

3.5 Conclusion

The musculoskeletal injury in wheelchair fencing is unique in nature. Despite the similarities in rules, skills and equipment, the injury patterns of wheelchair fencers are greatly different from those of able-bodied fencers. The findings from this pilot study showed wheelchair fencers have higher risk to sustain from upper limb extremity injuries. Wheelchair fencers with compromised trunk control have even higher risk in severe upper limb injuries than those with good trunk control. Following van Mechelen's (1992) injury prevention model (Figure 2.4); identifying prevalence and severity of sport injury, the next step would be conducting biomechanical evaluation to inaugurate the aetiology and mechanism of injury; so that an evident, practical and rational prevention program could be formulated.

Note: Part of this chapter was published in Clinical Journal of Sports Medicine 2012; 22(3): 278-280 and presented in The 3rd Student Conference on Sports Medicine, Rehabilitation and Exercise Science, Hong Kong, 19 June, 2010.

CHAPTER 4

Pilot study (1): Repeatability of kinematic and electromyographic data during lunge attack in wheelchair fencing

4.1 Introduction

Information obtained from three-dimensional (3D) kinematic studies can be used in identifying possible sports injury mechanism, assisting training progress and monitoring rehabilitation improvements, thus enhancing athletes' performance. However, using 3D kinematic to study upper limb motion had been challenging due to the large degrees of motion freedom, complex multi-joint structure and unconstrained cyclical movements of upper limbs. The inertial effect and oscillation of external markers over the underlying skeletal or soft tissue structures, especially during rapid and jerky motions may also affect accuracy of measurement. Possible inaccuracy may be related to closely positioned markers, changing inter-marker distances or simplification of arm into simple joints in some systems (Anglin & Wyss, 2000; Pearcy, et al., 1987). Due to such challenges, most upper limb 3D kinematic motion studies thus far had only investigated simple joint movements (Rab, et al., 2002; Rau, et al., 2000; Reid, et al., 2010; Mackey, et al., 2005). With the development of more sophisticated motion tracking system, and advance in the calibration of the recording system, some of these errors could be minimized. Additionally, more meticulous marker placement could also ensure repeatability and visibility throughout the test (Anglin & Wyss, 2000).

Establishing reliability of the application of 3D kinematic technology to analyze upper limb motions is important to ensure it is a valid research tool to produce consistent and reproducible measures for assessment, diagnosis as well as monitoring change due to intervention (Fagarasanu & Kumar, 2002).

There were a number of validity and reliability research on 3D motion analysis, included patients with brachial plexus palsy and children with hemiplegia (Lempereur, et al., 2012; Rab, et al., 2002; Reid, et al., 2010; Mackey, et al., 2005). All these investigations, however, were related to sterotyped movements and slow in action. These obviously will be different from sports that require complex movement patterns and demands quick and explosive actions. More importantly, the reliability and validity of using 3D motion analysis on wheelchair fencing actions has not been conducted.

Similarly, surface electromyography (SEMG) is a common method used to analyze neuromuscular functions. However, the reliability of SEMG measurement during wheelchair fencing had not been explored. Since the main research study in this dissertation focused on the kinematic and EMG measurements of wheelchair fencing in a single session, it is essential to examine the within-session repeatability of the 3D upper limb kinematics and EMG variables. The purpose of this study is to evaluate the reliability of the 3D optical kinematic and surface EMG variables during lunge attack in wheelchair fencing. It was hypothesized that kinematic and EMG values recorded by the current protocol could yield high within-session reliability. Meanwhile, this pilot project also served to examine the technical choice of the experimental setup and the safety issue during the lunge attack motion in wheelchair fencing.

4.2 Method

4.2.1 Participants

Ten experienced male wheelchair foil fencers from the Hong Kong Paralympic Wheelchair Fencing Team voluntarily participated in the study. The demographic characteristics of the participants are summarized in Table 4.1. All participants were screened for the absence of recent injury or major operation done to their fencing arm or spine. Written consent was signed after the details of this study were explained to them (Appendix V and VI). The study had been approved by the ethical review committee of the Hong Kong Polytechnic University.

Table 4.1 Demographic characteristics of the wheelchair fencers (n=10)

Subject	Age (year)	Height (m)	Weight (kg)	Fencing experience (year)	Fencing arm	Disability
1	27	1.79	80.2	7	Left	Above knee amputation
2	38	1.75	68.5	12	Left	Poliomyelitis
3	32	1.73	65.5	10	Right	Spinal cord injury T10
4	42	1.68	69.3	5	Right	Hemiplegia
5	39	1.73	55.0	11	Right	Spinal cord injury T4
6	24	1.72	50.5	5	Left	Spinal cord injury T6
7	20	1.83	68.0	4	Left	Poliomyelitis
8	23	1.77	55.9	5	Right	Below knee amputation
9	30	1.75	65.3	8	Left	Poliomyelitis
10	36	1.82	65.0	9	Right	Congenital limb deficiency
Mean	31.11	1.76	64.32	7.50	Right: 5	
SD	7.53	0.05	8.56	2.99	Left: 5	

4.2.2 Data collection

4.2.2.1 Kinematic recording

4.2.2.1.1 Kinematic model

The upper extremity model consists of the five following segments: (1) trunk, (2) right upper arm, (3) right forearm, (4) left upper arm, and (5) left forearm. The segments are connected by a t hree degree-of-freedom (DoF) shoulder joint (glenohumeral joint), a two DoF elbow joint and a two DoF wrist joint. The three DoF in the shoulder can be attributed to abduction-adduction, flexion-extension and external-internal rotation of humerus relative to the scapula (the shoulder joint here refers only to the glenohumeral joint). Two DoF in the elbow joint correspond to pronation-supination and flexion-extension, while two DoF in the wrist joint correspond to flexion-extension and radial-ulnar deviation. Twenty-four reflective markers (10-mm spheres) were attached to each participant according to the Vicon Upper Limb Model Product Guide Vision 1.0. O xford Metrics Ltd, Oxford, UK (2007). The landmarks for reflective markers attachment were bilateral acromio-clavicular joints, lateral epicondyles, medial epicondyles, radial styloids, medial styloids, distal third metacarpus, 7th cervical vertebra, 10th thoracic vertebra, jugular notch and xiphoid process. Upper Limb marker descriptions were shown in Table 4.2. Another eight markers for technical reference frame were attached to bilateral upper arm (3 markers for each side) and forearm (1 marker each for each side) respectively (Figure 4.1). Six additional markers were placed on the foil to determine its orientation in space. Adhesive tape was applied to firmly secure each reflective marker to minimize the movement between markers and the underlying bony structures. A pilot trial showed that such arrangement could provide a secure marker attachment during the briskly lunge attack motion.

Subject specific anthropometric measurements are recorded by the investigators of this study and fed in the model as the subject parameters. Those parameters together with the marker coordinates are used to define the centers and axes of joints. Vicon BodyBuilder V3.55 (Vicon Motion Systems, Ltd., Oxford, UK) was used for the development of the model. Euler angles are used to determine the 3D joint angles (Grood & Suntay, 1983; Ramakrishnan & Kadaba, 1991; Harris & Smith, 1996). A series of Euler rotations, sequenced Y-Z-X, were used to express the joint angles of the distal segment with respect to the proximal segment by utilizing each segment's local coordinate system. The trunk segment was described with reference to the lab coordinate system. Fencing lunge movements were captured with 8 MX-40 cameras at a sampling rate of 200 Hz.

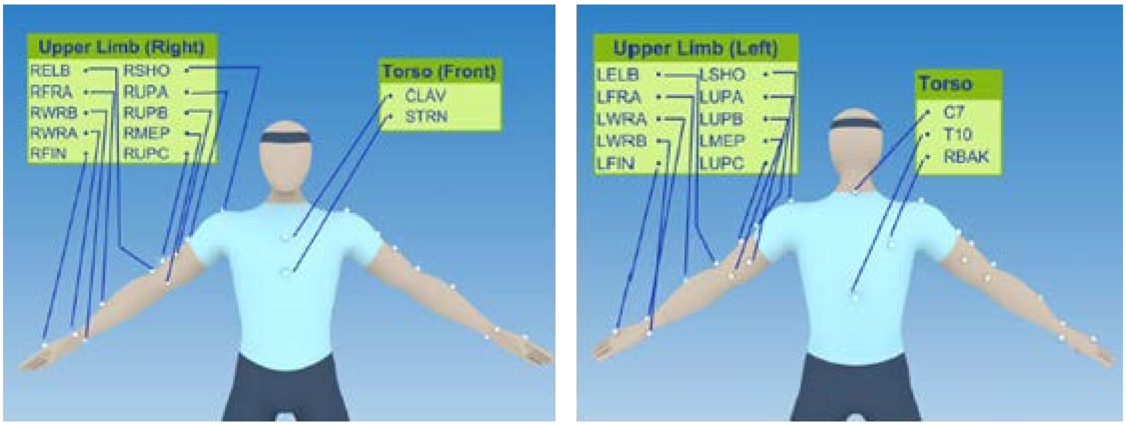


Figure 4.1 Marker sets attachment in Upper Limb Model (left: front view; right: rear view) (Vicon Upper Limb Model Product Guide Version 1.0, 2007)

Table 4.2 Description of marker placement (Vicon – Upper Limb Model Product Guide Version 1, 2007)

Marker	Definition	Marker Placement
C7	7th cervical vertebra	On the spinous process of the 7th cervical vertebra
T10	10th thoracic vertebra	On the spinous process of the 10th thoracic vertebra
CLAV	Clavicle	On the jugular notch where the clavicles meet the sternum
STRN	Sternum	On the xiphoid process of the sternum
LSHO	Left shoulder	On the acromio-clavicular joint
LUPA	Left upper arm marker A	On the lateral upper left arm
LUPB	Left upper arm marker B	On the lateral upper left arm
LCPC	Left upper arm marker C	On the lateral upper left arm
LELB	Left elbow	On the lateral epicondyle approximating the elbow joint axis
LMEP	Left humerus medial epicondyle	On the left humerus medial epicondyle
LFRA	Left forearm	On the lateral left forearm
LWRA	Left wrist marker A	At left radial styloid attached symmetrically with a wristband on the posterior of the left wrist
LWRB	Left wrist marker B	At left ulnar styloid attached symmetrically with a wristband on the posterior of the left wrist
LFIN	Left finger	Just below the left third metacarpus
RSHO	Right shoulder	On the acromio-clavicular joint
RUPA	Right upper arm marker A	On the lateral upper right arm
RUPB	Right upper arm marker B	On the lateral upper right arm
RCPC	Right upper arm marker C	On the lateral upper right arm
RELB	Right elbow	On the lateral epicondyle approximating the elbow joint axis
RMEP	Right humerus medial epicondyle	On the right humerus medial epicondyle
RFRA	Right forearm	On the lateral right forearm
RWRA	Right wrist marker A	At right radial styloid attached symmetrically with a wristband on the posterior of the right wrist
RWRB	Right wrist marker B	At right ulnar styloid attached symmetrically with a wristband on the posterior of the right wrist
RFIN	Right finger	Just below the right third metacarpus

4.2.2.1.2 Joint center

The joint center locus of the shoulder, elbow and wrist had to be firstly determined to define the model. The position of the joint center served as the origin for each segment's local coordinate system and all joints were assumed to have fixed centers of rotation (Figure 4.2). Details of method to determine the joint center loci were listed as follow:

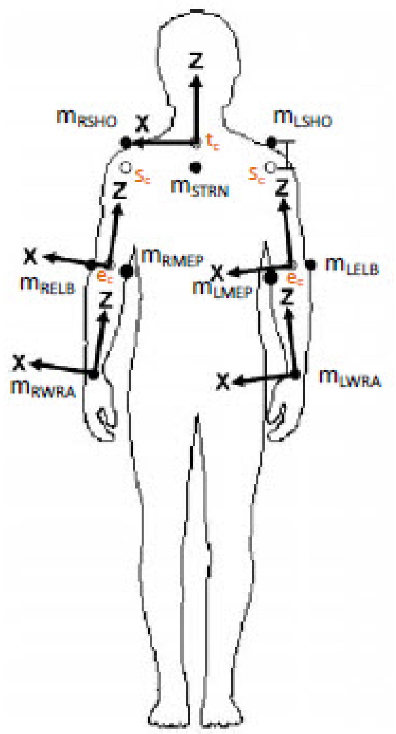


Figure 4.2 Local coordinate axes systems for the upper extremity model. Black and open markers indicate the corresponding marker positions and joint centers respectively. Axes follow the convention of X: flexion/extension, Y: abduction/adduction and Z: axial rotation

Shoulder: The glenohumeral joint was modeled as a ball and socket joint, with no translation of the rotation center of the humerus. The joint center was located at the center of the humeral head (Wang, et al., 1998; Biryukova, et al., 2000; Bachschmidt, et al., 2001). The shoulder thickness was measured by caliper (d) to determine joint center location. The joint center was located inferior to the acromion, at the measured distance of d/2 and was defined as follows:

$$\overline{s}_c = m_{SHO} - (d/2)(\overline{t}_z)$$

where $\overline{s}_c$ was the 3D location of the shoulder center, m_{SHO} was the 3D location of the acromion marker and $\overline{t}_z$ was the z-axis unit vector of the trunk coordinate system.

Elbow: The elbow joint center was assumed to lie anterior to the olecranon process and the half way between the lateral and medial epicondyle.

$$\overline{e}_c = \frac{1}{2}(\overline{m}_{ELB} + \overline{m}_{MEP})$$

where $\overline{e}_c$ is the center of the elbow; $\overline{m}_{ELB}$ and $\overline{m}_{MEP}$ were the 3D location of medial and lateral epicondyle markers respectively.

Wrist: The wrist joint center was located halfway between the radial and ulnar styloid processes:

$$\overline{w}_c = \frac{1}{2}(\overline{m}_{WRA} + \overline{m}_{WRB})$$

where $\overline{w}_c$ was the center of the wrist; $\overline{m}_{WRA}$ and $\overline{m}_{WRB}$ were the 3D location of radial and ulnar styloid process markers respectively.

4.2.2.2 EMG recording

The electromyographic signals of eight muscles over the fencing arm during lunge attacks were collected using surface electromyographic method. The eight muscles were upper trapezius (UT), infraspinatus (INF), anterior deltoid (ANT), mid-deltoid (MID), long head of biceps brachii (BIC), long head of the triceps brachii (TRI), wrist flexors (WF) and wrist extensors (WE).

Double-differential Ag-AgCl surface electrodes with an inter-electrode distance of 10 mm (DE-3.1, Delsys Inc, USA) were used. The surface electrode was adhered to the target muscles according to the guidelines of SENIAM project (Surface Electromyography for a Non-Invasive Assessment of Muscles, BIOMED II, 1997) (Figure 4.3). A reference electrode was placed over the acromion process of the non-fencing arm. The skin of each electrode site was shaved, cleaned with alcohol swab and lightly grazed with fine sandpaper to reduce skin impedance. To minimize movement artifact during the rapid lunge action, all electrodes and leads were fixed to arm and trunk of the tested subjects by adhesive tapes.

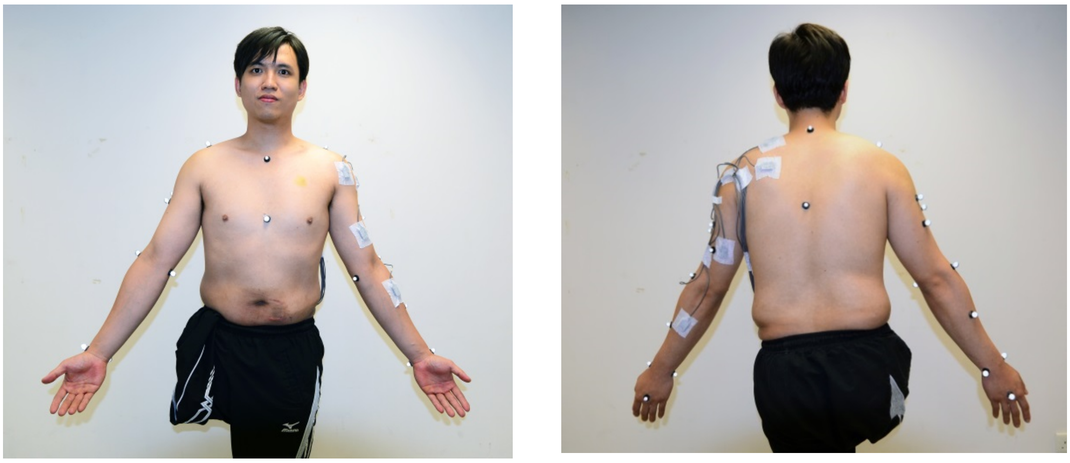


Figure 4.3 EMG electrode placements (left: front view; right: rear view)

4.2.2.3 Testing procedures

4.2.2.3.1 Experimental setup

4.2.2.3.1.1 Cameras

Four wall-mounted cameras and four cameras fixed on tripods were positioned around the tested subjects (Figure 4.4). All reflective markers were inspected via the workstation to ensure their visibility during five testing trials of fencing lunge action toward the hitting dummy. After positioning of the cameras, the Vicon system was calibrated with standard procedure and to ensure the calibrated residual was less than 0.3 as recommended by the manufacturer. The system accuracy was within 0.5 mm in XYZ axes by calibration trial.

Figure 4.4 Optoelectrical camera placement and setup (left: front view; right: rear view)

4.2.2.3.1.2 Wheelchair

A standard wheelchair was clamped firmly on a wheelchair platform that was used in official wheelchair fencing competition. The wheelchair platform was anchored to the ground to ensure the wheelchair fencers' safety during the impulsive lunge action (Figure 4.5).

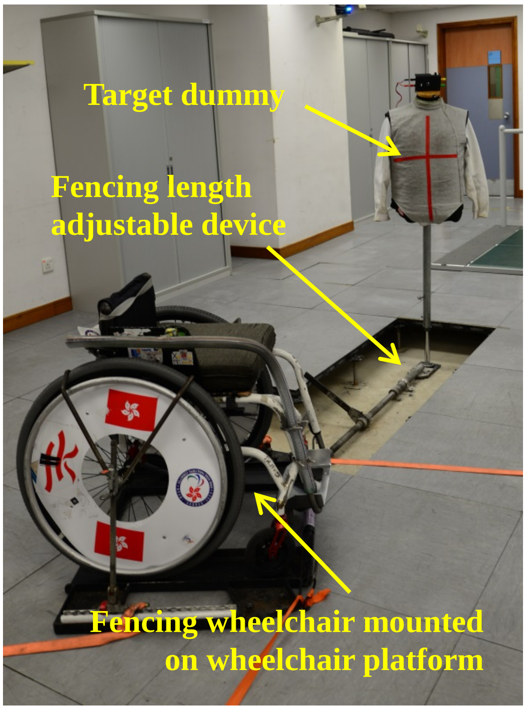


Figure 4.5 Fencing wheelchair, target dummy and fencing length adjustable device

4.2.2.3.1.3 Target dummy and fencing length adjustable device

A dummy hitting target was fixed on a length adjustable device that was mounted on the floor (Figure 4.5). A light was attached to the upper part of dummy. The light was control by the investigator of this study; when the light was on, it signaled the fencer to start the lunge attack. A fencing jacket was put onto the dummy and connected to the Vicon system. Once the fencer's foil hit the target, an electrical signal could be sent to the Vicon system to indicate the completion of lunge attack.

The lunge hitting targets were placed at four different zones on t he dummy to simulate the actual fencing scenarios. The four zones were the right-upper (RU), right-lower (RL), left-upper (LU) and left-lower (LL) (Figure 4.6).

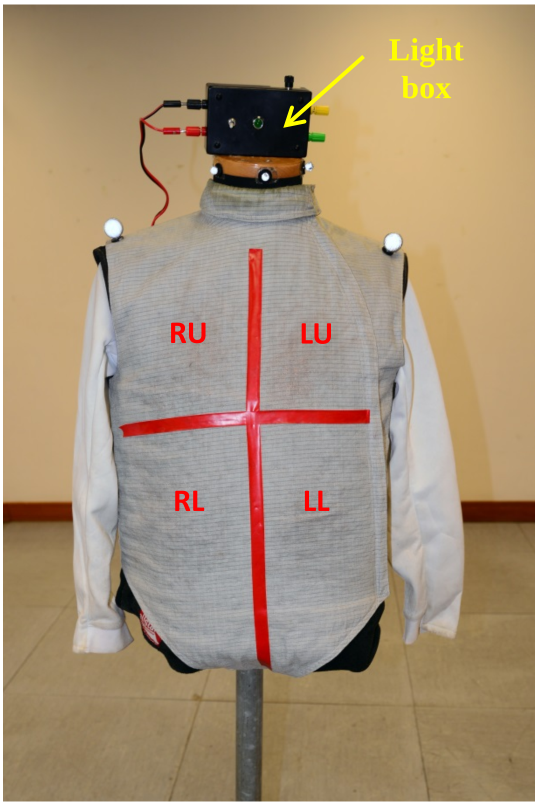


Figure 4.6 Target areas on dummy (RU: Right Upper, LU: Left Upper, RL: Right Lower and LL: Left Lower). A light source was fixed on the front and top part of dummy

4.2.2.3.2 Maximal voluntary isometric contraction (MVIC) recording

Before collecting the fencing kinematic and EMG data, SEMG activity of each target upper limb muscle during maximal voluntary isometric contraction (MVIC) was obtained. Standardized MVIC protocols to isolate the activation of individual muscle based on standard muscle testing techniques were used (Table 4.3). After skin preparation, EMG electrodes were affixed onto the fencers' upper limbs and trunk with double-side adhesive tape. Subjects were instructed to perform three 5-second

MVIC trials for each target muscle against a dynamometer held by the investigator. The participants were verbally encouraged to use maximum effort throughout the 5-second contraction period. Readings from dynamometer were recorded and counterchecked to ensure the subject had consistently contracted their muscles in the successive MVIC testing. One-minute rest period was given between each MVIC trial. Only the middle 3 seconds of the MVIC data were utilized to compute the peak EMG values to avoid the effect of unsteady muscle work at the initial phase of muscle contraction and fatigue at the end of isometric contraction task. Mean value of the three repetitions was used to represent the MVIC EMG.

Table 4.3 Testing positions for 8 upper limb muscles over the fencing arm (Kendall, 1993)

Muscle	Testing position
Upper trapezius	30° shoulder abduction, downward force applied to upper arm
Infraspinatus	90° elbow flexion, internally rotated force applied to forearm
Anterior Deltoideus	45° shoulder flexion, downward force applied to elbow
Medius Deltoideus	90° shoulder abduction with forward force applied to the wrist
Biceps brachii	90° elbow flexion, full supination, downward force applied to wrist
Triceps brachii	90° shoulder abduction, full internal rotation, 45^0 elbow flexion, downward force applied to wrist
Wrist Flexors	90° elbow flexion, full pronation, downward force applied to dorsum of wrist
Wrist Extensors	90° elbow flexion, full supination, downward force applied to palmar aspect of wrist

4.2.2.3.3 Wheelchair fencing lunge motion

The key focus of this study was to examine the fencing arm motion during lunge attack - the fundamental and most frequently used motion in wheelchair fencing. It involves quick, brisk and highly-coordinated motion of the upper limb and torso. According to the competition rule of the International Wheelchair Fencing Committee (IWFC), a v alid lunge attack requires wheelchair fencer's buttock remains in full contact with the seat-cushion during the fencing actions (IWAS, 2013). Under such restrictions, even if fencers had partial lower extremity functions, the hip and leg motion was minimal. An experienced wheelchair fencer was present on-site to judge the technical validity of the lunge attacks performed by each participant. Only technically valid lunge attacks were included in the data analysis of this study.

After the MVIC recording, the subjects were asked to take a 5 minutes rest. During the rest time, reflective markers were then attached to the body landmarks of each participant. Participants were seated in a s tandard wheelchair, which was fastened to a rigid metal frame. The target dummy was adjusted to the shoulder height of the subject for height standardization amongst individuals. Fencers were asked to perform the lunge attack to a dummy using a foil at normalized fencing distance (Distance_100). The normalized fencing distance is the official starting position in wheelchair fencing and is measured as the length of the foil plus one full arm length of the tested fencer and the dummy arm length (Figure 4.7). In the present study, the dummy arm length was pre-determined and set at 23 cm. Fencers initiated lunge attack when the signal light on the target started flashing. The lunge action was completed when the foil hit the electrical scoring apparatus (Figure 4.8). Each fencer repeated 5 trials of lunge attack to the targets at their fastest speed, with a 30-second rest between trials. A 1-minute rest was given to the fencer before proceeding to the next target zone. To minimize the learning effect, the order of lunge attacks to each target zone was randomized by drawing lots. Electrical light source signals and the outputs of the scoring detection apparatus were synchronized with the Vicon analogue input channels and the EMG capturing systems. The study was performed at the Motion Laboratory of the Hong Kong Polytechnic University.

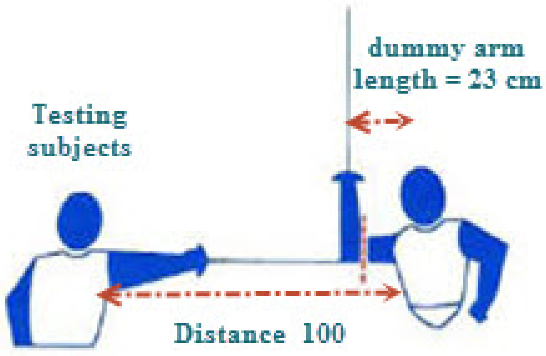


Figure 4.7 Normalized fencing distance (Distance_100), IWFC, 2011

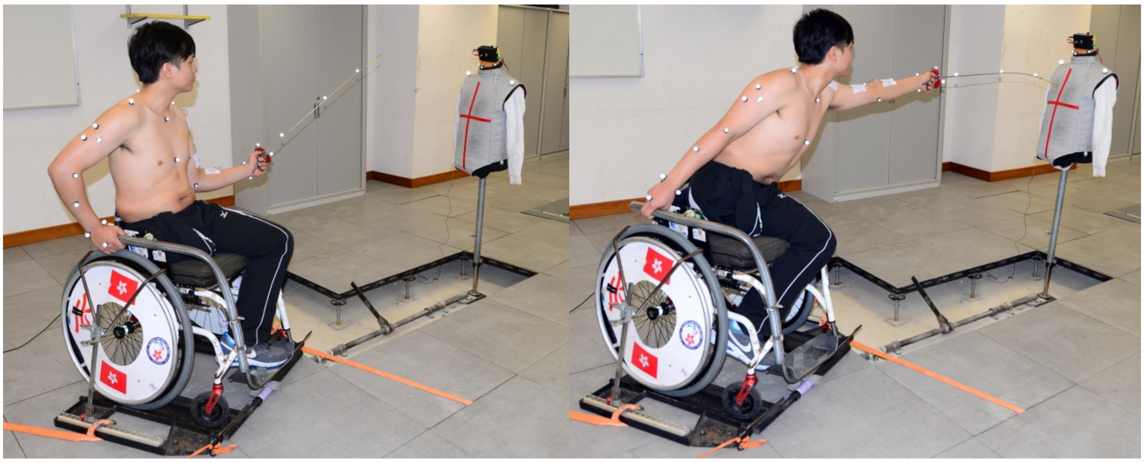


Figure 4.8 Lunge attack motion at Distance_100 as performed by a Category A fencer (left: starting position; right: finishing position)

4.2.3 Data reduction

4.2.3.1 Kinematic data

Data from the Vicon cameras were tracked, reconstructed and processed by the Upper Limb Plug-in-Gait Model and Vicon Nexus 1.6 Software (Oxford Metrics Inc, Oxford, UK) to extract the angular displacement data (in ASCII format) of the shoulder, elbow and wrist joints of the fencing arm. The angular displacement data was smoothed by Butterworth 2nd order filter (Low-pass filter at 8Hz) before computation and then pipelined to Microsoft Excel (Microsoft Corp., Redmond, WA, USA) for processing. Angular displacement-time history data of the fencing arm joints for each lunge trial was further time normalized to 100 data points using a cubic spline function within the customized MatLab codes (The MathWorks Inc, Natick, MA, USA) for movement pattern comparison, Appendix VII. All data for left-handed fencers were converted to read as data for right-handed fencers.

4.2.3.2 EMG data

All SEMG signals were pre-amplified by 1000 times; differentially amplified with common mode rejection ratio (CMRR) >80 dB and filtered with a bandwidth of 20-450 Hz (Bagnoli-4 Main Amplifier, DelSys Inc., USA) to remove the possible movement artifacts. The signals were digitized by an A/D converter (PCI-6033E, National Instrument Inc, USA) with a sampling rate of 1000 Hz. Electrical signals from light source and scoring apparatus were output to the electromyographic systems via an analogue-to-digital converter, and displayed graphically as a voltage signal for identifying the start and end of lunge attack motion. Raw SEMG data were processed offline using custom-made MATLAB program (Appendix VIII) and Microsoft Excel 2007 (Microsoft, Corp, Redmond, Washington) to generate the integrated-EMG (iEMG) and peak EMG values. To examine the repeatability of the EMG testing, raw SEMG signals were full wave rectified and smoothed at 10 Hz to produce linear envelopes for each muscle. SEMG signal for each lunge attack cycle was time-normalized to 100 data points using a cubic spline function. EMG amplitude variable of the peak EMG value was normalized against the MVIC EMG value for each tested muscle.

4.2.4 Statistical analysis

Descriptive statistics of the mean angular displacement of shoulder, elbow and wrist joint; and normalized peak EMG and iEMG at the four fencing targets were collected and analyzed. The Normal Gaussian distribution of all kinematic and EMG data were ascertained by the Shapiro-Wilk test. One-way repeated measures analysis of variance (ANOVA) was used to examine any differences amongst the four fencing targets. Any significant differences detected were followed by post-hoc Tukey's honestly significant test and paired t-test with Bonferroni correction.

Intra-class correlations ICC $_{(3,1)}$, based on the two-way mixed model consistency type, was employed to ascertain the test-retest reliability of the mean angular displacement and peak normalized EMG values over the five trials (Portney & Watkins, 2009; Shrout & Fleiss, 1979). ICC coefficients less than 0.50 represents poor reliability, 0.50 to 0.75 suggests moderate reliability, and above 0.75 indicates good reliability (Portney & Watkin, 2009).

The repeatability of the normalized angle-time waveforms and SEMG linear envelopes obtained by the five trials was determined by the coefficient of multiple correlation (CMC), using the following formula (Kadaba, et al., 1989; Steinwender, et al., 2000; Lee, et al., 2003).

$$\sqrt{1-\frac{\sum_{i=1}^{5}\sum_{j=1}^{n}(A_{ij\_}\bar{A}_j)^2/5(n-1)}{\sum_{i=1}^{5}\sum_{j=1}^{n}(A_{ij\_}\bar{A})^2/(5n-1)}}$$

where $A_{ij\_}$ is the jth sample point of the ith set of the angle or EMG data, $\bar{A}_j$ is the mean at the jth sample point over the five data sets, and $\bar{A}$ is the grand mean over the n sample points and the five data sets. The numerator of the right-hand side of Equation is the variance of the waveform data about a 'running' mean ($\bar{A}j$) or the ensemble mean curve across the five data sets, and the denominator is the variance about the grand mean (Lee, et al., 2003). CMC value of 1 indicates high repeatability of waveforms and 0 indicates low repeatability. SPSS version 19.0 (SPSS Inc., USA) and MedCalc version 9 software packages were used for statistical analysis. A significant level was set at 0.05 for all the statistical analyses.

4.3 Results

4.3.1 Angular displacement, peak EMG and integrated EMG

The mean angular displacements and EMG parameters of the fencing arm during lunge at the four fencing targets are summarized in Table 4.4, 4.5 and 4.6. No significant difference was detected among the kinematic variables of the four fencing targets except for the forearm pronation (p=0.004). There was no significant difference found for all peak EMG and integrated EMG data at the four fencing targets (p>0.05).

Table 4.4 Angular displacement (degree) of the upper limb joint movement during lunge attack.

Joint movement	Fencing target				
	RU	RL	LU	LL	P values
Shoulder abduction	70.7±8.6	68.8±12.1	77.1±10.1	71.2±4.8	$F_{3,27}$=2.578; p=0.120
Shoulder flexion	96.4±16.5	96.7±9.3	95.4±8.8	97.9±8.5	$F_{3,27}$=0.154; p=0.926
Shoulder internal rotation	60.9±26.5	61.7±24.6	60.2±19.8	63.3±18.0	$F_{3,27}$=0.494; p=0.689
Elbow extension	66.9±8.7	62.8±3.6	64.6±5.3	61.1±3.4	$F_{3,27}$=1.848; p=0.162
Forearm pronation*	28.6±7.8	19.6±7.9	29.8±13.5	13.0±2.1	$F_{3,27}$=10.169; p=0.004
Wrist flexion	13.1±2.1	14.1±2.5	13.5±3.7	12.5±2.5	$F_{3,27}$=0.558, p=0.647
Wrist ulnar deviation	11.3±2.1	11.0±2.2	8.4±1.8	10.6±2.0	$F_{3,27}$=5.016, p=0.700

LL: Left lower, LU: Left upper, RL: Right lower and RU: Right upper; *p<0.05

Table 4.5 Peak EMG values (%MVIC) of the eight fencing arm muscles during lunge at different fencing target for 10 experienced wheelchair fencers

Peak EMG	Fencing target				
	RU	RL	LU	LL	P values
Upper Trapezius	21.7±6.4	23.4±6.4	22.1±7.6	20.5±8.0	$F_{3,27}$=1.218; p=0.322
Infraspinatus	20.2±6.6	20.4±5.8	22.1±7.7	21.1±4.0	$F_{3,27}$=0.561; p=0.645
Anterior deltoid	34.1±5.8	34.9±6.5	32.9±3.4	31.5±4.1	$F_{3,27}$=0.879; p=0.464
Mid detoid	19.7±3.9	20.1±2.3	20.0±2.1	19.3±2.5	$F_{3,27}$=0.267; p=0.848
Biceps	25.7±8.9	24.7±10.9	22.1±5.2	24.9±8.9	$F_{3,27}$=0.719; p=0.550
Triceps	8.0±2.4	8.4±2.3	8.3±1.7	9.5±3.9	$F_{3,27}$=1.232; p=0.317
Wrist flexors	5.7±1.5	7.0±2.5	6.7±2.8	8.1±2.8	$F_{3,27}$=0.799; p=0.505
Wrist extensors	9.3±2.9	12.0±2.6	11.4±3.7	10.9±2.1	$F_{3,27}$=2.145; p=0.118

LL: Left lower, LU: Left upper, RL: Right lower and RU: Right upper; *p<0.05

Table 4.6 Integrated EMG values (%MVIC) of the eight fencing arm muscles during lunge at different fencing target for 10 experienced wheelchair fencers

Integrated EMG	Fencing target				
	RU	RL	LU	LL	P values
Upper Trapezius	23.1±9.2	24.2±9.0	20.7±9.8	22.7±9.9	$F_{3,27}$=1.734; p=0.184
Infraspinatus	31.8±14.2	31.1±10.4	30.5±13.4	31.2±9.7	$F_{3,27}$=0.135; p=0.938
Anterior deltoid	43.8±8.6	48.1±8.2	42.0±8.6	46.3±8.1	$F_{3,27}$=1.484; p=0.241
Mid deltoid	25.3±5.1	27.4±2.6	24.0±3.2	26.6±3.7	$F_{3,27}$=1.925; p=0.149
Biceps	27.9±7.6	29.7±9.0	25.6±6.7	26.6±7.1	$F_{3,27}$=1.927; p=0.149
Triceps	8.4±4.1	8.8±3.8	7.4±3.8	9.3±4.4	$F_{3,27}$=1.861; p=0.200
Wrist flexors	4.8±0.9	4.8±1.8	4.2±1.6	6.0±4.1	$F_{3,27}$=1.616; p=0.209
Wrist extensors	14.0±4.3	16.2±4.7	15.4±4.8	15.1±5.7	$F_{3,27}$=1.664; p=0.198

LL: Left lower, LU: Left upper, RL: Right lower and RU: Right upper, *p<0.05

4.3.2 Movement pattern comparison

A typical movement waveform was depicted by all fencers as shown in Figure 4.9. To prepare for the lunge position, fencer kept his shoulder in an abducted, flexed and externally-rotated position. The elbow was flexed with forearm slightly supinated, wrist extended and radial-deviated. From this initial position, fencers started off the lunge by initiating shoulder abduction, flexion and internal rotation, followed by elbow extension to advance the weapon towards the target. When the weapon approached the target, fencers pronated their forearms with wrists flexed and ulnar-deviated to control the foil to hit onto the target.

The mean ICC and CMC values for angular displacement ranged from 0.73-0.98 and 0.70-0.98 respectively (Table 4.7). When examined the ICC and mean CMC values of the angular displacement at the four fencing targets, a h igh repeatability estimates were generally detected (RU – ICC: 0.76-0.95, CMC: 0.70-0.98; RL – ICC: 0.73-0.96, CMC: 0.71-0.93; LU – ICC: 0.79-0.97, CMC: 0.71-0.95; LL – ICC: 0.84-0.98, CMC: 0.70-0.92). The mean CMC values were slightly lower for shoulder rotation (0.73 ± 0.20) and forearm supination (0.70 ± 0.22).

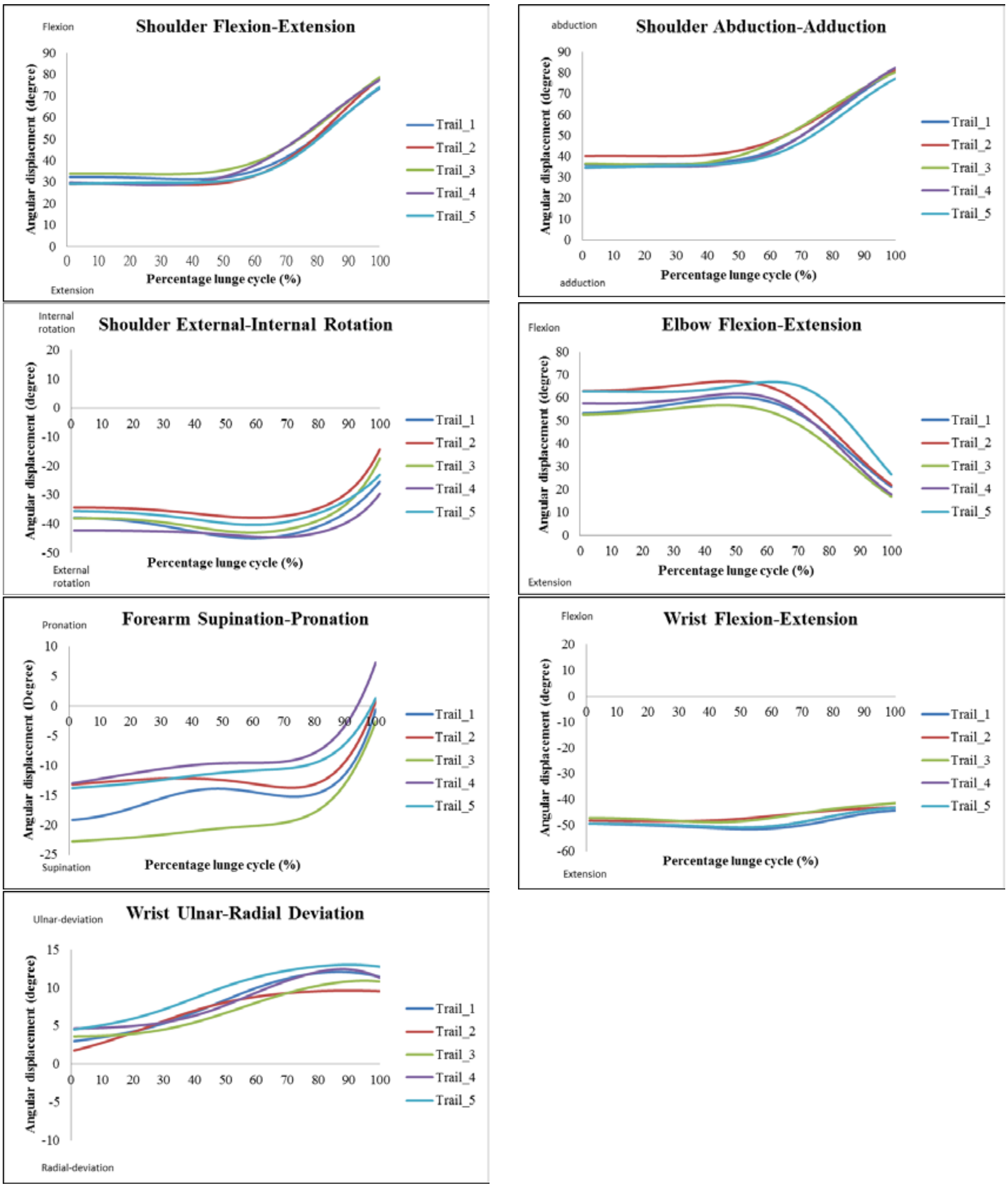


Figure 4.9 Waveforms showing the movement pattern of the upper limb joint motions during the 5 trails of lunge attack (degrees) as performed by a representative wheelchair fencer

Table 4.7 Mean coefficient of multiple correlation (CMC) and intraclass correlation coefficient ($ICC_{3,1}$) for within-session upper limb kinematic measurement during lunge action of fencing

	CMC ± standard deviation				$ICC_{3,1}$ (95% CI) for peak angular displacement			
	RU	RL	LU	LL	RU	RL	LU	LL
Shoulder flexion	0.97 ± 0.01	0.87 ± 0.03	0.95 ± 0.01	0.92 ± 0.03	0.88 (0.50-0.97)	0.91 (0.61-0.98)	0.87 (0.42-0.94)	0.98 (0.88-0.99)
Shoulder abduction	0.98 ± 0.01	0.93 ± 0.02	0.90 ± 0.02	0.88 ± 0.01	0.81 (0.51-0.95)	0.89 (0.54-0.97)	0.91 (0.60-0.98)	0.87 (0.42-0.99)
Shoulder internal rotation	0.73 ± 0.20	0.71 ± 0.10	0.73 ± 0.05	0.72 ± 0.04	0.95 (0.74-0.99)	0.96 (0.81-0.99)	0.96 (0.80-0.99)	0.92 (0.95-0.97)
Elbow extension	0.94 ± 0.05	0.92 ± 0.03	0.94 ± 0.05	0.86 ± 0.05	0.92 (0.64-0.98)	0.95 (0.73-0.99)	0.97 (0.84-0.99)	0.97 (0.84-0.99)
Forearm supination	0.70 ± 0.22	0.73 ± 0.09	0.71 ± 0.02	0.70 ± 0.22	0.95 (0.77-0.99)	0.95 (0.74-0.99)	0.89 (0.55-0.97)	0.89 (0.55-0.99)
Wrist extension	0.74 ± 0.15	0.72 ± 0.06	0.73 ± 0.05	0.75 ± 0.08	0.77 (0.29-0.93)	0.88 (0.51-0.97)	0.84 (0.42-0.95)	0.84 (0.42-0.95)
Wrist ulnar deviation	0.78 ± 0.05	0.74 ± 0.02	0.76 ± 0.04	0.78 ± 0.05	0.76 (0.60-0.99)	0.73 (0.22-0.91)	0.79 (0.54-0.97)	0.89 (0.54-0.97)

RU: Right Upper, RL: Right Lower, LU: Left Upper and LL: Left Lower

4.3.3 EMG pattern comparison

A consistent EMG pattern was observed amongst all wheelchair fencers. The proximal muscles including upper trapezius, infraspinatus, deltoid and biceps exerted a higher muscle activity during the lunge attack motion (Figure 4.10). Qualitative analysis also revealed that upper limb muscles followed the proximal-to-distal activation sequence. The mean CMC of the eight fencing arm muscles ranged from 0.70-0.94 (Table 4.7). The mean ICCs of the peak EMG and iEMG values ranged from 0.62-0.93 and 0.72-0.98 respectively (Table 4.8). CMC of the eight fencing arm muscles and EMG at the four fencing targets were generally high (RU: 0.73-0.94, RL: 0.70-0.92, LU: 0.71-0.91 and LL: 0.73-0.85). High ICC estimates for both the peak- (RU: 0.73-0.90, RL: 0.69-0.92, LU: 0.64-0.93 and LL: 0.69-0.95) and the integrated-EMG (RU: 0.76-0.98, RL: 0.72-0.98, LU: 0.79-0.97 and LL: 0.78-0.97) data of distal upper limb muscles were also found to be moderate to high for the four fencing targets.

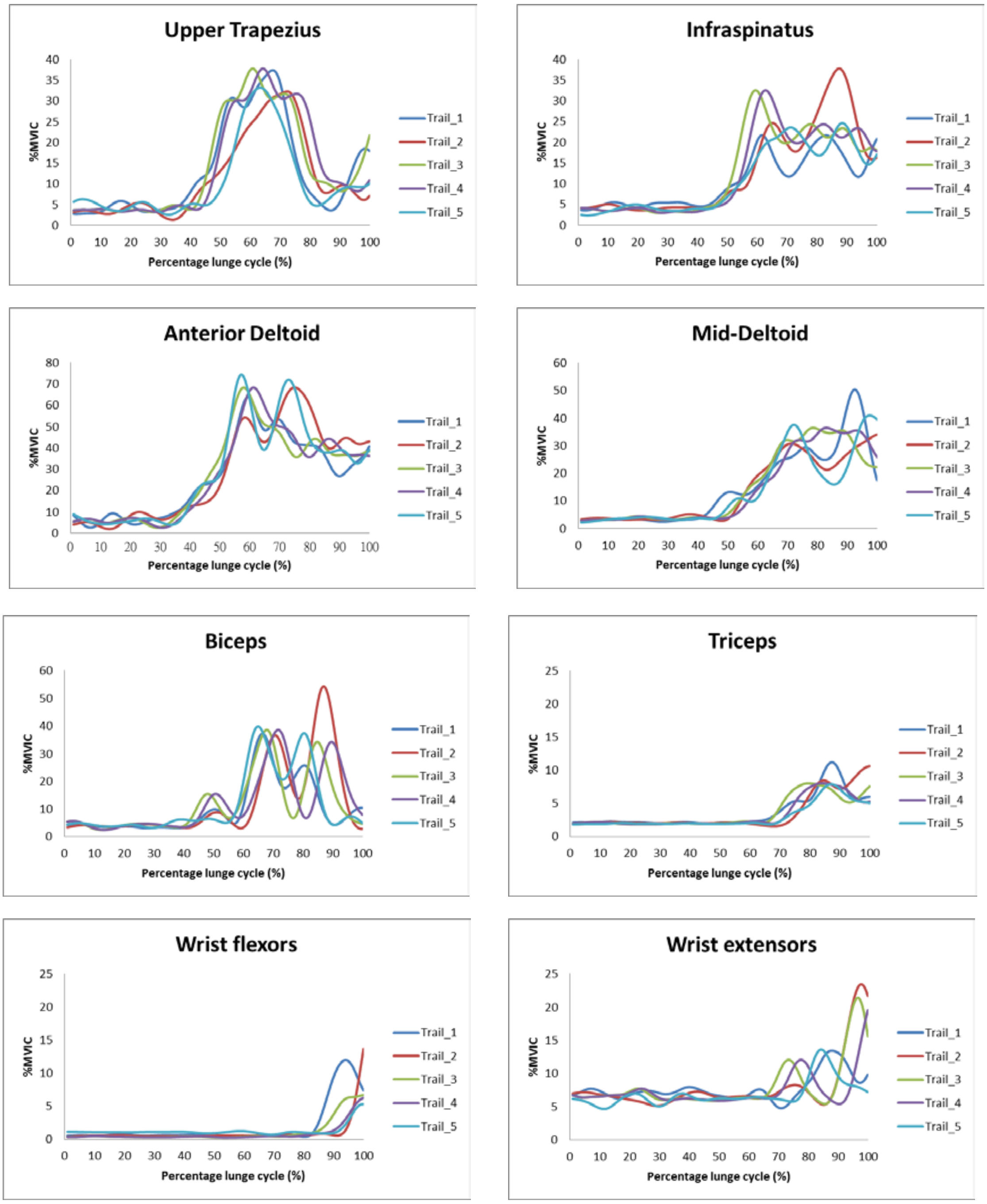


Figure 4.10 EMG envelopes showing the electromyographic pattern of the eight fencing arm muscles by a representative wheelchair fencer during the 5 trails of lunge attack

Table 4.8 Mean coefficient of multiple correlation (CMC) and intraclass correlation coefficient ($ICC_{3,1}$) of the Peak EMG and Integrated-EMG values of the eight upper limb muscles during lunge attack

	CMC (standard deviation)				$ICC_{3,1}$ (95% CI) for Peak EMG				$ICC_{3,1}$ (95% CI) for Integrated EMG			
	RU	RL	LU	LL	RU	RL	LU	LL	RU	RL	LU	LL
Upper Trapezius	0.86 (0.06)	0.81 (0.07)	0.83 (0.04)	0.85 (0.03)	0.79 (0.58-0.93)	0.89 (0.72-0.97)	0.72 (0.36-0.89)	0.77 (0.41-0.93)	0.96 (0.91-0.99)	0.94 (0.84-0.98)	0.90 (0.75-0.97)	0.87 (0.66-0.96)
Infraspinatus	0.94 (0.03)	0.90 (0.05)	0.80 (0.05)	0.85 (0.02)	0.90 (0.45-0.95)	0.92 (0.80-0.98)	0.93 (0.82-0.98)	0.95 (0.87-0.99)	0.98 (0.94-0.99)	0.98 (0.95-0.99)	0.97 (0.93-0.99)	0.97 (0.92-0.99)
Anterior Deltoid	0.94 (0.03)	0.92 (0.07)	0.91 (0.03)	0.81 (0.04)	0.77 (0.39-0.86)	0.75 (0.69-0.86)	0.64 (0.49-0.90)	0.83 (0.57-0.95)	0.76 (0.41-0.93)	0.89 (0.73-0.97)	0.79 (0.47-0.94)	0.78 (0.44-0.94)
Mid Deltoid	0.84 (0.02)	0.86 (0.03)	0.85 (0.03)	0.76 (0.03)	0.74 (0.35-0.93)	0.74 (0.35-0.93)	0.73 (0.36-0.90)	0.75 (0.38-0.87)	0.80 (0.55-0.86)	0.94 (0.86-0.98)	0.79 (0.76-0.89)	0.83 (0.63-0.87)
Biceps	0.84 (0.06)	0.81 (0.04)	0.85 (0.05)	0.78 (0.04)	0.86 (0.65-0.96)	0.85 (0.63-0.96)	0.87 (0.68-0.96)	0.89 (0.72-0.97)	0.93 (0.83-0.98)	0.87 (0.74-0.97)	0.89 (0.73-0.97)	0.88 (0.70-0.97)
Triceps	0.90 (0.03)	0.88 (0.11)	0.78 (0.07)	0.78 (0.05)	0.81 (0.52-0.95)	0.70 (0.57-0.86)	0.75 (0.38-0.93)	0.74 (0.35-0.93)	0.98 (0.96-0.99)	0.80 (0.51-0.94)	0.94 (0.85-0.98)	0.94 (0.85-0.98)
Wrist Flexors	0.75 (0.12)	0.70 (0.08)	0.71 (0.11)	0.73 (0.03)	0.76 (0.46-0.94)	0.70 (0.24-0.86)	0.72 (0.58-0.89)	0.69 (0.23-0.91)	0.78 (0.46-0.94)	0.83 (0.58-0.95)	0.88 (0.71-0.97)	0.88 (0.75-0.97)
Wrist Extensors	0.73 (0.04)	0.71 (0.03)	0.72 (0.04)	0.75 (0.03)	0.73 (0.33-0.92)	0.69 (0.46-0.76)	0.73 (0.56-0.87)	0.74 (0.24-0.82)	0.80 (0.60-0.94)	0.72 (0.31-0.92)	0.83 (0.83-0.98)	0.89 (0.72-0.97)

RU: Right Upper, RL: Right Lower, LU: Left Upper and LL: Left Lower

4.4 Discussion

A consistent lunge attack pattern involving shoulder flexion / abduction / internal rotation, elbow extension, forearm pronation and wrist flexion / ulnar-deviation was identified. The high EMG activities of supraspinatus, infraspinatus, deltoids and biceps have highlighted their major role in lunge attack. The angular displacement, peak EMG and integrated EMG values of the four hitting targets (i.e. RU, RL, LU and LL) at normalized fencing distance (i.e. Distance_100) showed no significant difference. Further, the high CMC and ICC showed that the current optical kinematic analysis and surface EMG measurements generated excellent within-session repeatable data.

4.4.1 Choice of fencing target for the main study

Similar to able-bodied fencing, wheelchair fencing has official scoring areas depending on the types of fencing weapons used (i.e. foil, epee and sabre). In foil, the valid scoring area only involves the torso region (Harmer, 2008). To systematically examine the fencing performance of the participants in the current study, the official trunk scoring area was divided into four hitting target zones (i.e. right upper (RU), right lower (RL), left upper (LU) and left lower (LL)). The secondary objective of this reliability study was to assess the feasibility of the experimental setup in the subsequent experiments; the findings of this study refined the experimental protocol used in the main study.

Preliminary finding showed that there was no significant difference in the upper limb joint angular displacement, peak EMG and iEMG of the fencers during their lunge attack towards the four hitting targets zones (Table 4.4, 4.5 and 4.6). The overall reliability estimates for kinematic and EMG during the lunge attacks at all targets were high (Table 4.7 and 4.8), indicating all the kinematic and kinetic analysis of lunge attack at all targets were similar. A small focus group was also conducted with the wheelchair fencers to collect feedback on their preferred target zone. Several wheelchair fencers raised concerns about the risk of falling from wheelchair when hitting the LU and LL targets. Considering in the main study, the fencing distance would increase up to 15% of the normalized fencing distance, both LU and LL targets were discarded from the main study to minimize the risk of fall.

Approximately over 80% of the participated wheelchair fencers prefer hitting their opponent's right upper trunk during real competition; as attacking the opponent's right lower trunk will expose the fencer's upper trunk to opponent's attack. The RU was selected as the hitting target in the subsequent experiments in this thesis.

4.4.2 Repeatability of the kinematic measurement

The current study result showed high within-session repeatability and is comparable to other previous 3D upper limb studies despite the different study samples used. Jaspers, et al. (2011) found high within-session reliability (ICC >0.6) for all upper limb joint angles during reaching, reach-to-grasp and upper limb gross motor tasks in healthy children. The within-session reliability (CMC) for shoulder and elbow joint angles in cerebral palsy (CP) and typically developing (TD) children during side reach (CP: 0.63 to 0.82; TD: 0.65 to 0.72) and hand-to-month tasks (CP: 0.54 to 0.94; TD: 0.19 to 0.84) were also reported high (Reid, 2010). The high mean ICC (0.80) and CMC (0.70) for angular displacement indicates the wheelchair fencers performed consistent lunge attacks and the current study procedures are also reliable for capturing the kinematic during the rapid lunge attack motion.

Intra-subject variation must be minimized to ensure reliability of the 3D upper limb kinematic measurements. Due to the limited lower limb physiological joint movements, there are only limited motor strategies that can be used to complete a task and such motor strategies are usually highly-repeatable (Bergmann, et al., 2009; Bronner, et al., 2010; Kadaba, et al., 1989). The possible motor strategies could be used for upper limbs to complete a task are significantly more due to the available joint movements (Rau, et al., 2000). No standard activities exist for the arm; the same motor task could be achieved by different strategies (Hingtgen, et al., 2006; Mackey, et al., 2005; Reid, et al., 2010). Bernstein (1967) described the anatomical redundancy present in the upper limb as the 'degrees of freedom' or 'motor equivalence' problem. The large degrees of freedom in the upper limb allow an infinite number of joint angles, leading to the large variation of movement patterns in upper limb kinematic measurement (Grea, et al., 2000; Robertson & Miall, 1997). In the present study, the repeatability for movements in both sagittal and frontal planes was higher than those in the transverse plane. This difference may be due to the fact that lunge execution mainly occurs in the sagittal and frontal planes. Moreover, the participants were elite fencers; their skill levels might have allows them to easily reproduce unique lunge attack motion that reduced the variations of the kinematic data between trials.

Errors related to movement artefact are usually due to the anchorage of external markers during the fast lunge action. Markers mounted on skin to measure the position and movement of the underlying skeleton may impose some errors, particularly in the shoulder complex and elbow. The wide range of motion as well as high degrees of freedom of the shoulder complex may impose challenges in the use of skin markers; while humeral epicondyle markers are difficult to mount on skin due to the definition of elbow joint (Reid, et al., 2010). Rotations of long-axis extremity included upper arm internal-external rotation and forearm pronation-supination are very sensitive to skin movement measurement artefacts (Roux, et al., 2000). Additionally, any briskly, sudden motion may cause considerable oscillation of the reflective markers, thus may reduce the repeatability of the kinematic measurement. Extensive precautions were taken in this study to minimize all these potential sources of errors. First, the markers were secured by double-sided adhesive tapes. Second, reflective marker placement procedure was performed by an experienced physiotherapist who accurately locate the bony landmarks for the Vicon model. Third, eight cameras were used simultaneously and were all positioned to provide multiple viewpoints to effectively solve marker obstruction (Wang, et al., 2003). Lastly, a high sampling frequency of 200 Hz for motion capture was selected in this study, following the learnings from other previous upper limb kinematic studies that involved fast motion such as baseball (Fleisig, et al., 1999) and able-bodied fencing (Frere, et al., 2011). All these measures together were found to be effective based on the high reliability indexes achieved in present study.

The high CMC and ICC values substantiate the use of the Vicon system for 3D kinematic analysis for rapid and powerful lunge attack motion. The repeatability of the distal upper limb joints movement was slightly lower than the proximal region. The relatively lower repeatability could be explained due to the lower absolute range of motion (ROM) of distal joints. The absolute ROM values of the distal joint such as forearm and wrist during lunge attack were much smaller than those of the shoulder and elbow joints (Table 4.4). Other previous studies also found when measuring joints with the limited absolute ROM, every degree difference in the repeated measurements would inevitably magnify the variance of the distal joint movement (Mackey, et al., 2005; Reid, et al., 2010; Steinwender, et al., 2000); any small errors coming from skin movement would significantly lowered the repeatability of the distal joint that has small displacement patterns (Kadaba, et al., 1989).

4.4.3 Repeatability of the EMG measurement

Previous studies that involved powerful muscular activities such as vertical jump, running and taekwondo kick generally produced poor to fair reliability of the EMG variables (Aggeloussis, 2007; Goodwin, 1999; Karamanidis, 2004). A number of factors could influence surface EMG measurement included the protocol selection, size and type of electrode and electrode placement (Aggeloussis, 2007; Deluca, 1997; Farina, et al., 2004; Goodwin, et al., 1999; Karamanidis, et al., 2004; Kollmitzer, et al., 1999; Rota, et al., 2013). Depending on the type, magnitude, velocity and synchronization of the muscle contractions of the impulsive powerful dynamic tasks, movement of muscle relative to skin could happen and result in the change of contact face of the muscle under the electrodes, thus inconsistency of EMG measurements (Rota, et al., 2013). However, our study obtained highly CMC and ICC values despite the quick, briskly and jerky lunge attack motions; supporting the use of SEMG testing with our experimental setup and testing protocol for the analysis of wheelchair fencing biomechanics.

Some previous motion analysis studies showed the repeatability of the proximal muscles EMG were less than the distal muscles. Kadaba & coworkers (1989) showed that the within-in session coefficient of variation (CV) values were lower for leg muscles gastrocnemius (53±8%) and anterior tibialis (49±6%) as compared to hamstrings (62±11%), va stus lateralis (56±8%), adductor longus (63±11%) and gluteus maximus (56±9%) during self-pace walking. Similar findings by Winter and Yack (1987) showed the within-session CV values of gastrocnemius (33±9%), anterior tibialis (33±5%), hamstrings (62±21%), vastus lateralis (46±26%), adductor longus

(41±10%) and gluteus maximus (42±13%). These studies all showed a trend of reduced reliability from distal to proximal muscles; probably due to two reasons: the relatively larger size of muscles and higher degrees of freedom of the proximal muscle groups. The proximal muscles are generally larger in size and more active during the lunge execution. The possible change in shape of muscle during the forceful contraction may alter the electrode contact as well as the received electric signals for repeatable measurement. The upper limb joints, particularly the glenohumeral joint, have higher degrees of freedom; potentially allowing a large variation of motor patterns across the five trials of lunge attack motion. However, this study yielded similar CMC and ICC values for all selected muscles surrounding the shoulder joints (i.e. upper trapezius, infraspinatus and deltoids). The high performance level of our wheelchair fencers and the standardization of our instructions to all fencers for ensuring maximal work during lunge attack may contribute to the small intrasubject variation for the lunge technique performed.

The mean CMC values of electromyographic were generally lower than those of kinematics. Although the reason for this difference is yet to be clear, similar observation was reported by Kadaba, et al. (1989).

4.4.4 Limitations

This study tested the intra-tester reliability. The inter-tester study yet has to be conducted despite our experimental setup and protocol should likely be reproduced by other trained examiners with comparable experience. Furthermore, other factors including the motivation of the subjects and the re-application of tracking systems, that may affect the repeatability of our kinematical and EMG measurements, were not addressed in this study.

4.5 Conclusion

Our results support both the optical tracking and surface EMG methods are reliable and technically feasible tools to examine the upper limb motion for lunge attack in wheelchair fencing. The experimental 3D kinematic and surface EMG protocol for the fencing arm during lunge attack motion developed in our laboratory was highly repeatable.

Note: Part of this chapter was presented in The Hong Kong Physiotherapy Association Conference, Hong Kong, 22-23 November, 2011.

CHAPTER 5

Pilot study (2): Validity of the optical tracking method for fast upper limb kinematic measurement

5.1 Introduction

Chapter 4 has demonstrated that the optical tracking method has high repeatability in capturing three-dimensional (3D) kinematic data during the fast fencing lunge action. This chapter reports a pilot study compared the rapid arm movement kinematic measured by an optical tracking system against an acceptable current bench mark - an inertial tracking system (Saber-Sheikh, et al., 2010; Thies, et al., 2007). The purpose of this pilot study was to establish the validity of the optical tracking for upper limb rapid movements. Inertial tracking systems are commonly used for kinematic measurements in the field of biomechanics and sports medicine in virtue of their light-weight, portability, affordability and easy application (Luinge, et al., 1999; Luinge & Veltink, 2005; Sabatini, et al., 2005). Most of the existing studies compared the agreement of inertial and optical tracking systems were restricted to measuring lower limb motions or slow arm movements (Bergman, et al., 2009; Thies, et al., 2007; Zhou, et al., 2006; 2008), their results could not be generalized to rapid arm motion analysis. Accordingly, this study aimed to compare the agreement of the upper limb joint angle measurements during high speed motions between inertial and optical tracking systems.

5.2 Methods

5.2.1 Participants

Thirty healthy male volunteers (aged 25.1 ± 3.2 years; height 178.6 ± 5.3 cm) participated in this study. Individuals with recent upper limb injuries or operations were excluded. This study had been approved by the Ethics Review Board of the Hong Kong Polytechnic University and the participants provided written consents prior to participation (Appendix IX and X).

5.2.2 Instruments

The two systems used in this study were the Vicon Motion analysis system for optical tracking and Xsens MTx sensors system for inertial tracking.

5.2.2.1 Optical motion analysis system

The Vicon Motion Analysis System (Vicon MX, VICON, Oxford, UK) with a matrix of 8 wall-mounted 200 Hz cameras was used to track the positions of the optical reflective markers in 3D space. Twenty four retro-reflective 10 mm diameter markers were attached onto the participants' dominant upper limb and trunk according to the Upper Limb Model of the VICON Upper Limb Model Product Guide Version 1.0., Oxford Metrics Ltd, Oxford, UK (2007) as described in Section 4.2.2. The Vicon Nexus 1.6 Software and Upper Limb Plug-in-Gait Model (Oxford Metrics Inc, Oxford, UK) was used to capture and analyze the upper limb motion.

5.2.2.2 Inertial tracking system

Xsens MTx sensors system (Xsens Technologies; Enschede, Netherlands) was used to measure 3D acceleration, 3D rate of turn and magnetic field. The system incorporates 3D gyroscopes, accelerometers and magnetometer which were reported to provide drift-free motion data (Saber-Sheikh, et al., 2010). The sensors (38 x 53 x 21 mm, 30g) transmitted digital signal to Xbus master system and connected to the personal computer to calculate three degrees of freedom orientation data at a sampling rate of 200 Hz (Figure 5.1). The sensors were attached at four landmarks: 1) the third thoracic vertebral level, 2) the middle part of an upper arm, 3) the proximal forearm, and 4) the dorsal aspect of the dominant hand. All sensors were secured by straps to improve sensors fixation (Figure 5.2). The manufacturers reported static accuracy for roll/pitch and yaw was <0.5 degree and <1 degree respectively while the dynamic accuracy was 2 degree root mean square (MTi and MTx user manual and technical documentation, 2010).

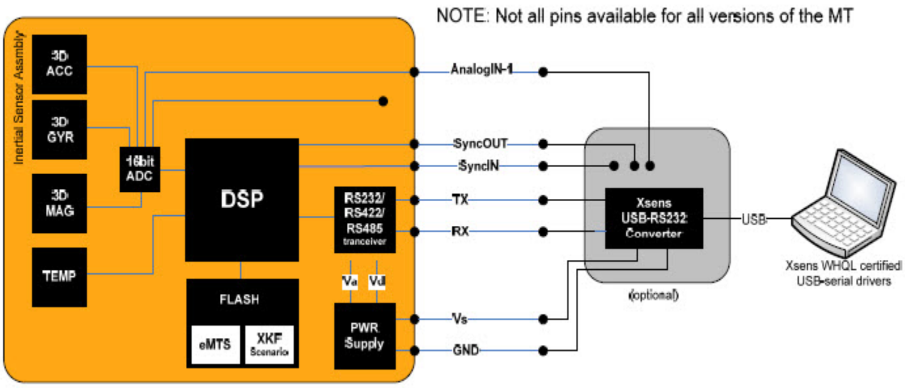


Figure 5.1 Xsens MTx sensors system overview

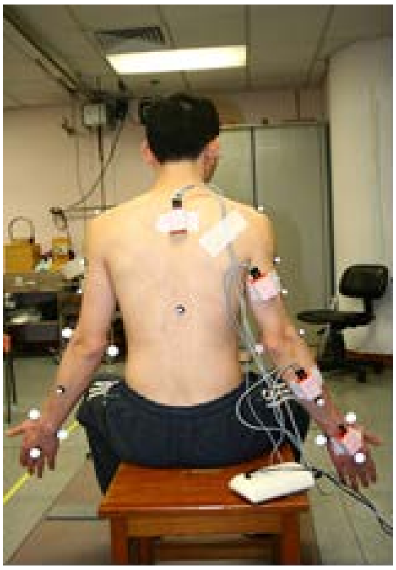


Figure 5.2 Xsens MTx sensors placement

5.2.3 Testing procedures

After markers and sensors were attached, participants were instructed to move their dominant arm in six specific motions in random order at their maximum speeds: shoulder flexion, shoulder abduction, shoulder external rotation, elbow flexion, forearm supination and wrist flexion in sitting position. The motions were captured simultaneously by the Vicon and Xsens systems. Each joint movement was repeated five times with 1-minute rest interspersed between trials. A pulse signal, captured by one of the Vicon analog channels, was used to synchronize the Xsens data.

5.2.4 Data analysis

The 3D angular displacements of the tested shoulder, elbow and wrist joints by the optical tracking were derived from Upper Limb Plug-in-Gait Model and Nexus 1.6 Software (Oxford Metrics Inc, Oxford, UK). Data acquisitions for inertial sensors were performed using MT Manager Version 1.7.0 configured for human motion, as suggested by the manufacturer. The drift-free 3D orientation data were corrected by mathematical procedures established in previous studies (Saber-Sheikh, et al., 2010; Woltring, 1994). Further data processing was performed with Matlab (The MathWorks Inc, Natick, MA, USA). A second order Butterworth low-pass with cut-off frequency of 8Hz was applied and kinematic outputs for the two systems were modeled by the same Eurler angle decomposition sequence (Doorenbosch, et al., 2003; Wu, et al., 2005). Angular displacement of shoulder (flexion, abduction and external rotation), elbow (flexion) and wrist (flexion) for each trial obtained by the two tracking methods were compared and the results were averaged for statistical analysis. Angular displacement-time history data for each movement trial was also time normalized to 100 points (represented 0-100% of the movement cycle) using a cubic spline function for movement pattern comparison.

5.2.5. Statistical analysis

The kinematic data of each joint as measured by the two systems at the inner, middle and outer range for each joint motion were compared using two-tailed Pearson product-moment correlation coefficients (r) to evaluate the strength of relationship between the angular displacements. The differences of the means and the 95% confidence limits of the two methods were graphically examined using the 95% limits of agreement (Bland & Altman, 1986; 2010). The similarity of motion curves of each of the six specific motions between the two tracking methods were evaluated by the coefficient of multiple correlation (CMC); CMC ≥ 0.7 indicates acceptable similarity of continuous kinematics between repeated trials (Kadaba, et al., 1989). All data were analyzed using MedCalc 11.3.5 (MedCalc Software bvba; MedCalc, Mariakerke, Belgium) and SPSS version 18 (SPSS Inc., Chicago, IL, USA) software packages, with the statistical significance level for all tests set at 0.05.

5.3 Results

The Pearson's correlation coefficients of the six specific motions were statistically significant ($p<0.01$). Specifically, the Pearson's correlation coefficients for shoulder, elbow and wrist joint motion ranged from 0.71 to 0.87, from 0.78 to 0.99, and from 0.76 to 0.84 respectively (Table 5.1). The average peak angular velocity of shoulder, elbow and wrist motions are shown in Table 5.1.

All joint angles measured by the two systems were within the 95% limits of agreement (Appendix XI). The mean difference between the Vicon and Xsens measurements ranged from 0.45° to 1.05° (Table 5.1). CMCs of the arm movement waveforms ranged from 0.75 to 0.95 (Table 5.1). The particular motions showed excellent CMCs ($p<0.01$) are shoulder flexion ($r=0.92$), shoulder abduction ($r=0.91$) and elbow flexion ($r=0.95$).

5.4 Discussion

Results of the present study demonstrated high correlation and agreements between the optical motion analysis system and the inertial tracking system in 3D analysis of high velocity upper limb motion. The Bland and Altman plots (Appendix XI) illustrated that there was neither a systematic error associated with joint angles nor any significant systematic difference in joint angle measurements between the two systems. These results concurred with previous studies that the optical tracking method and inertial sensors for upper limb kinematic measurements have high agreement (Roetenberg, et al., 2007; Zhou & Hu, 2007; 2010) and supported that the two systems are interchangeable for rapid arm motion analysis.

Apart from high agreement between optical and inertial tracking systems, our findings also showed that the inertial tracking system was accurate for high-speed upper limb motion analysis. Previously, inertial tracking system was mainly studied for slow limb movements. There was concern that rapid motion might affect the accuracy of motion detectors (Forner-Cordero, et al., 2008; Mayagoitia, et al., 2002) and recommended using algorithms to correct the inertial sensor-related upper limb measurement errors (Zhou, et al., 2008). However, our study did not apply any mathematical adjustment, and the two systems demonstrated high measurement agreement for high-speed arm motion analysis. We believe the protocol and experimental set up contributed to the success of the results. Extensive precautions were taken to ensure optical markers were not obstructed from cameras by positioning the participants in the centre of the capture zone, and closely monitored any occlusion during the data collection. We also secured the inertial sensors to the arm with double-sided adhesive tapes and straps to minimize movement artefact.

The angular velocities of upper limb joint movements in this study were found to be comparable to those observed in the wheelchair fencing pilot study covered in the next chapter. The current findings support the use of either tracking methods in wheelchair fencing research. However, the optical system has several advantages over the inertial method in this study. First, the cables connecting the inertial sensors restrict the natural fencing motions. Second, the size and weight of the inertial sensors may hinder normal joint movements, especially to smaller distal joints. Third, the preload compression due to the application of extra straps in securing the inertial sensors may induce unnecessary discomfort to the wheelchair fencers. Last but most importantly, inertial sensors and surface electromyography electrodes compete for the common spots on the upper limb (e.g. triceps and wrist extensors), which made the simultaneous kinematic and electromyographic data capture impossible. Considering these technical issues, the optical tracking method appears to be more suitable for the motion analysis of wheelchair fencing and was thus selected in the subsequent research study in this dissertation.

This study was limited by the investigation of single-planar movements of individual arm joints separately; our results may not be generalized to multi-planar and multi-joint motion analysis. However, given our standard measurement procedures, our protocol is very likely to yield high measurement agreements for multi-planar motion analysis.

5.5 Conclusion

This study demonstrated high correlation and agreements between the optical and inertial tracking systems for rapid upper limb motion analysis and substantiate their use for 3D rapid arm motion analysis such as the lunge action in wheelchair fencing. The optical tracking system is preferable for wheelchair fencing kinematic analysis because no attachment of sensors to the fencing arm is needed.

Note: Part of this chapter was presented in The Hong Kong Physiotherapy Association Conference, Hong Kong, November 22-23, 2011.

Table 5.1 Correlations and 95% limits of agreement of Xsens and Vicon joint angle measurements

Joint motion	Angular velocity ± SD (°/s)	Predetermined angles	Mean difference (degree)	95 % CI (degree)	95% limits of agreement(degree)		Pearson's correlation coefficients	CMC	
					upper limit	lower limit		Mean	SD
Shoulder flexion	880.3 ± 81.3	30° 90° 150°	-0.07° -0.78° -0.61°	-0.11 to -0.02 -0.97 to -0.59 -0.78 to -0.44	0.17 0.20 0.28	-0.30 -1.76 -1.50	R = 0.87 R = 0.85 R = 0.79	0.92	0.02
Shoulder Abduction	900.2 ± 89.7	30° 90° 150°	-0.39° -0.43° -0.03°	-0.45 to -0.34 -0.51 to -0.34 -0.12 to 0.05	-0.11 -0.01 -0.40	-0.67 -0.86 -0.47	R = 0.80 R = 0.81 R = 0.72	0.91	0.03
Shoulder rotation	870.6 ± 102.3	20° 40° 60°	-0.40° -0.32° -0.32°	-0.50 to -0.30 -0.41 to -0.22 -0.44 to -0.20	0.13 0.20 0.29	-0.92 -0.83 -0.93	R = 0.71 R = 0.77 R = 0.72	0.75	0.21
Elbow flexion	972.2 ± 88.6	30° 90° 120°	-0.27° -0.54° -0.38°	-0.40 to -0.13 -0.62 to -0.45 -0.52 to -0.23	0.45 -0.07 0.39	-0.99 -1.00 -1.14	R = 0.96 R = 0.97 R = 0.99	0.95	0.02
Forearm supination	1435.0 ± 107.3	30° 90°	-0.14° -0.47°	-0.31 to 0.04 -0.61 to -0.32	0.78 0.30	-1.05 -1.24	R = 0.78 R = 0.78	0.86	0.07
Wrist flexion	1019.2 ± 142.3	20° 40°	-0.36° -0.06°	-0.47 to -0.25 -0.13 to -0.02	0.20 0.34	-0.92 -0.45	R = 0.76 R = 0.84	0.85	0.06

CI: confidence interval; CMC: coefficient of multiple correlation between the Xsens and the Vicon systems; SD: standard deviation

CHAPTER 6

Kinematic and electromyographic analysis of fencing lunge attack in world-class wheelchair fencers

6.1 Introduction

Chapter 3 described the high prevalence of upper limb injuries in both Category A and B elite wheelchair fencers. With further compromise of trunk work, Category B fencers were shown to have even higher risk and more severe upper limb injuries that lead to prolonged absence from sport participation as a consequency. Bahr & Krosshaug (2005) suggested incorporating biomechanical analysis in the investigation of the causative injury mechanism for sport injury in order to formulate a rational and effective injury prevention program. In sports science and medicine, kinematic and electromyographic analyses are commonly used in biomechanical studies. While the kinematic data provide detailed description of movements, the EMG variables provide information about the motor pattern and level of muscular activities. Several biomechanical studies were conducted in able-bodied fencing and advocated the importance of footwork for the generation of powerful lunge attack (Morris, et al., 2011; Suchanowski, et al., 2011). In contrast, with footwork being eliminated, wheelchair fencers rely solely on their trunk and upper limb to execute all fencing motions (Fung, et al., 2013). For those with poor trunk control, their upper limbs may bear extra loading. Previous research had documented that individuals with poor trunk control demonstrated altered kinematics and increased EMG activities in wheelchair propulsion (Dubowsky, et al., 2009), daily transfer tasks (Gagnon, et al., 2008), forward-reaching movement (Seelen, et al., 1997) and wheelchair sport activities (Nunome, et al., 2002; Reid, et al., 2007). Based on the kinetic chain concept, the human body functions as a one kinetic unit, motion being transferred from foot to scapulohumeral joint and finally to the distal arm (Groppel, 1992). Any disruption of the kinetic chain may hinder the energy transfer and may cause higher risk of injuries. Applying this concept to wheelchair fencers, it is conceivable that the lack of footwork and trunk control may render the wheelchair fencers to sustain from a higher chance of upper extremity injuries. Quantitative measurements of the fencing motion kinematics, inter-joint coordination and muscular work would help to explore the possible injury mechanism in wheelchair fencing.

To date, research study on wheelchair fencing is scanty. Only one study had employed the two-dimensional video motion analysis method to investigate the trunk kinematic of the wheelchair fencers (Fung, et al., 2013). Three-dimensional upper limb kinematics and motor recruitment pattern studies during wheelchair fencing had not been conducted. Furthermore, despite the distinct difference in injury incidence between the wheelchair fencers with good- and poor- trunk control, no study were conducted to compare the kinematical and electromyographic differences between these two fencing groups. To bridge this research gap, this study aimed to measure and compare the three-dimensional (3D) upper limb kinematics and electromyographic activities of wheelchair fencers with good trunk control (Category A) and those with poor trunk control (Category B) during lunge attacks. We hypothesized that upper limb kinematics (lunge duration, angular displacement, peak linear velocity, peak angular velocity and inter-joint coordination) and surface EMG outcomes (time-domain: onset and occurrence of peak EMG, amplitude domain: peak EMG and integrated EMG, and inter-muscle coordination) of the 8 upper limb muscles over the fencing arm were different between Category A and Category B fencers at different fencing distances. Findings of the present study could provide insight for better understanding of the possible injury mechanism in wheelchair fencers, and more importantly; could facilitate the development of injury prevention program or rehabilitation protocol that is specific to wheelchair fencing.

6.2 Methods

Supported by the work described in Chapter 4 and 5, the repeatability, validity and practicability of the experimental protocol was established in this thesis. The two pilot studies indicated that both the optical motion analysis and EMG methods employed in the current study were valid and repeatable.

6.2.1 Participants

We invited all eligible male wheelchair fencers from both the Hong Kong and China Paralympic teams to participate in this study. All wheelchair fencers must be clinically diagnosed with permanent physical disability that resulted in substantial loss of one or both lower extremity function and must have received the International Wheelchair Fencing Committee (IWFC) permanent classification status. Currently there was no Category C wheelchair fencer in either the China or the Hong Kong Paralympic team. In order to minimize the variability of fencing skills for motion analysis, all recruited fencers must also be at elite level and had competed in any one of the Para-Asian Games, World-Championships and Paralympics. Fencers who had injured their upper limb during the previous 3 months or had suffered from any active pain or illness that impaired their fencing performance were excluded. Thirty male (15 Category A and 15 Category B) elite foil fencers were recruited. Twenty two of them were from the Hong Kong Paralympic Team and 8 of them from the China Paralympic Team. Three Category

A and four Category B recruited wheelchair fencers ranked among the top ten of the IWFC ranking list (2012) in their respective categories.

The demographic characteristics of the participants were summarized in Table 6.1 and there is no significant differences in the demographic profile between the two groups ($p>0.05$). The experimental procedures and the potential risks were explained to all participants prior to obtaining their written consents (Appendix XII and XIII). This study was approved by the ethical review committee of The Hong Kong Polytechnic University.

Table 6.1 Characteristics of fencers

Fencer group	Category A (n=15)	Category B (n=15)
Mean age (year)	30±8	32±12
Mean height (cm)	165±12	168±10
Mean weight (kg)	66±9	63±8
Year in fencing (year)	13±7	14±7
Training load (hour/week)	13.5±2.7	14.5±2.8
Disability	AKA (n = 3)	T3-4 Paraplegia (n=9)
	BKA (n = 3)	T5-6 Paraplegia (n=3)
	Poliomyelitis (n=4)	Poliomyelitis (n =3)
	L1 Paraplegia (n=5)	

AKA: Above Knee Amputation; BKA: Below Knee Amputation

T: Thoracic; L: Lumbar

6.2.2 Sample size calculation

There is no information regarding the change of upper limb kinematic variables and level of muscle activation during wheelchair fencing available in the literature to serve as reference. In order to estimate the sample size of the main study, a pilot study was conducted. Ten experienced wheelchair fencers (5 Category A and 5 Category B) were recruited and instructed to sit in a standardized wheelchair to perform 5 repetitions of lunge attacks at their highest speed. The target was set at the right-upper (RU) region of a dummy target located at the normalized fencing distance (i.e. Distance_100) and the details refer to Section 4.2.2.3.3. Similar testing procedures were repeated at three other fencing distances that were equaled to 105% (i.e. Distance_105), 110% (i.e. Distance_110) and 115% (i.e. Distance_115) of Distance_100. The angular displacement of different upper limb joints of the fencing arm was measured by the Vicon system while the corresponding EMG activities of various selected muscles were measured by surface EMG (see below for the detail procedure). The maximum difference in angular displacement among all upper limb joints between the two groups at different fencing distances were used to calculated the sample size. Similarly, the maximum difference in the peak EMG among all upper limb muscles between the two fencing groups at different fencing distances were used to calculate the sample size.

Findings in the pilot study revealed that the maximum difference in angular displacement between the two groups was found at shoulder abduction. Specifically, there were increases in shoulder abduction angular displacement for Category A

(Distance_100: 75.8 ± 3.4°, Distance_105: 71.9 ± 2.1°, Distance_110: 80.7 ± 4.1° and Distance_115: 85.0 ± 3.3°) and Category B (Distance_100: 69.5 ± 1.3°, Distance_105: 80.2 ± 6.2°, Distance_110: 90.7 ± 4.1° and Distance_115: 97.9 ± 4.0°) fencers with increasing fencing distance. Using an alpha of 0.05 and power = 0.8 for a two-tailed F-test of the interaction effect of repeated-measures ANOVA, the pilot data yielded an effective sample size of a minimum of 5 participants in each experimental group using the software G*Power, given the partial eta-square equal to 0.243, a correlation of 0.553 a mong repeated measurements and a non-sphericity correction of 0.371.

The pilot study revealed that the maximum difference in peak EMG between the two groups was found in upper trapezius. The peak EMG values of the upper trapezius increased in both Category A (Distance_100: 14.9 ± 2.9 %MVIC, Distance_105: 19.4 ± 2.9 %MVIC, Distance_110: 20.9 ± 1.3 %MVIC and Distance_115: 20.5 ± 5.3) and Category B fencers (Distance_100: 14.5 ± 2.2 %MVIC, Distance_105: 17.9 ± 2.6 %MVIC, Distance_110: 23.9 ± 1.7 %MVIC and Distance_115: 30.3 ± 8.6) as they performed lunge attack at longer fencing distances. The G*Power calculation based on the EMG data showed that a minimum of 7 participants in each experimental group was need for the experiment if the alpha value was set at 0.05 and statistical power at 0.8.

Together, these data indicated that a sample size with at least 7 subjects in each group was needed to reveal statistically significant differences between the two categories of wheelchair fencers across the four fencing distances if power was set at 0.8 and alpha set at 0.05.

6.2.3 Instrumentation

Fencing arm movements for the shoulder, elbow and wrist joints were recorded by an 8-camera optical motion analysis system at a sampling rate of 200 Hz (Vicon MX, VICON, Oxford, UK). Whereas, the surface EMG signals of the eight upper limb muscles including upper trapezius (UT), infraspinatus (INF), anterior deltoid (ANT), mid-deltoid (MID), long head of biceps brachii (BIC), long head of the triceps brachii (TRI), wrist flexors (WF) and wrist extensors (WE) were collected using the Electromyography System (DE-3.1, DelSys Inc, USA). These eight muscles are the prime movers and major stabilizers of the lunge actions. The setup of instrument and experiment procedures for optical motion and EMG system were identical to what had described in Section 4.2.2.

6.2.4 Experimental protocol

All fencers were given 5 minutes to warm-up after the study procedures were explained. After skin preparation, EMG electrodes were affixed onto the investigated muscles according to the guidelines of SENIAM project (Surface Electromyography for a Non-Invasive Assessment of Muscles, BIOMED II, 1997). A reference electrode was placed over the acromion process of the non-fencing arm. The fencers were then asked to perform the maximal voluntary isometric contraction (MVIC), by isometric contraction against manual resistance, of each investigated muscle group. Standardized MVIC protocols to isolate the activation of individual muscle based on standard muscle testing techniques were used as detailed in Section 4.2.2.3.1. After the MVIC recording procedures, reflective markers were attached to the bony landmarks of each fencer according to the Vicon Upper Limb Model Product Guide Vision 1.0 (2007). Fencers were then instructed to seat in a standard wheelchair for target dummy height adjustment and normalized fencing distance measurement, following the procedures that outlined in Section 4.2.2.3.3.

To simulate real competition scenarios of foil maneuver, fencers were asked to perform the lunge attack at four pre-determined distances in random order by drawing lots. These distances included the normalized fencing distance (Distance_100), and at 105%, 110% and 115% of normalized distance (Distance_105, Distance_110, and Distance_115 respectively). The target was set at the right-upper (RU) region of the dummy. Rationale for justifying the use of RU as the hitting target was listed in Section 4.4.1.

Fencers initiated the lunge when the signal light on the target dummy started flashing. The lunge action was completed as the foil hit the electrical scoring apparatus which was attached to the right-upper (RU) of the dummy. Each fencer performed 5 repetitive trials of lunge attack to the assigned target at their fastest speed. Five minutes rest was given to the fencer before proceeding to the next fencing distance.

To ensure strict compliance to the official rules for wheelchair fencing, an experienced wheelchair fencer was invited to govern the sitting position of each participant throughout the test - buttock must be in full contact with the seat-cushion during the fencing actions (IWAS, 2010). Under such restrictions, hip and leg motion was minimal; even if fencers had partial lower extremity functions. Only technical-valid lunge attacks were recorded and analyzed.

Kinematic and EMG data were reviewed immediately after testing to ensure proper data collection without marker dropout. The entire data collection session took approximately 2 hours.

6.2.5 Data acquisition

6.2.5.1 Kinematic data

Kinematic data of the shoulder, elbow and wrist joints of the fencing arm was processed by the Upper Limb Plug-in-Gait Model and Vicon Nexus 1.6 Software (Oxford Metrics Inc, Oxford, UK) as described in Section 4.2.3.1. A customized MATLAB codes (The MathWorks^{TM}, Natick MA, USA) was used for kinematic data processing (Appendix VII). All kinematic data was filtered and smoothed by Butterworth 2nd order filter (Low-pass filter at 8Hz). Kinematic variables consisted of lunge duration, angular displacement, peak linear horizontal, peak vertical velocity of the shoulder/elbow/wrist joints and peak angular velocity of the shoulder/elbow/wrist joint motions in the three anatomical planes.

Inter-joint coordination between shoulder and elbow, shoulder and wrist, and elbow and wrist during lunging action was examined by coefficient of cross-correlation (R). Peak correlation coefficient was used to test the strength of the correlation of kinematics between the two joints (Li & Caldwell, 1999; Shum, et al., 2007). Cross-correlation value ranged from -1 (out-of-phase) to 1 (in-phase). The peak correlation coefficient value of 1 means perfect correlation; while value greater than 0.8 is considered high correlation; indicating close coupling of movements (Shum, et al., 2007). Whereas, the phase association determines the time lag (expressed as % lunge cycle) for peak cross correlation. In the present analysis, the proximal joint was used as the reference, and a positive lag implies that the proximal joint moved earlier than the distal joint in the movement cycle (Shum, et al., 2005).

Kinematic variables were summarized and listed in Table 6.2.

Kinematic variables of the 5 lunge attack trials by each fencer were averaged. Means and standard deviation for each fencing group (i.e. Category A and Category B) were further calculated for analysis.

Table 6.2 Summary of all kinematic dependent variables

Kinematic variables (unit)	Joint / movement
Lunge duration (second: s)	-
Peak linear velocity (meter/second: m/s) • Horizontal • Vertical	Shoulder / Elbow / Wrist joints
Angular displacement (degree: °)	Shoulder • abduction-adduction • flexion-extension • internal-external rotation Elbow • flexion-extension • supination-pronation Wrist • Flexion-extension • radial-ulnar deviation
Peak angular velocity (degree/second °/s)	Shoulder • abduction-adduction • flexion-extension • internal-external rotation Elbow • flexion-extension • supination-pronation Wrist • flexion-extension • radial-ulnar deviation
Cross-correlation coefficient (R: no unit) Time lag (% lunge cycle)	Shoulder (flexion / abduction / rotation) - Elbow (extension / pronation) Shoulder (flexion / abduction / rotation) - Wrist (flexion / ulnar-deviation) Elbow (extension / pronation) - Wrist (flexion / ulnar-deviation)

6.2.5.2 EMG data

In this experiment, amplitude domain and time domain EMG variables were used to document the motor recruitment patterns associated with the 4 fencing distances and to compare the motor recruitment characteristic difference between the Category A and Category B fencers.

For amplitude domain EMG, the raw EMG signals were band-pass filtered (20-450 Hz), fully-waved rectified, smoothed with second-order Butterworth low pass filter (10Hz) and amplitude-normalized to MVIC EMG value for each investigated muscle group. Details refer to Section 4.2.3.2.

The determination of the SEMG onset was based on the instance that SEMG signal was at three standard deviations above the mean of the baseline value for 25 milliseconds (Beres-Jones, 2004; Dubowsky, et al., 2009). Temporal EMG data including the onset of SEMG burst and the occurrence of peak EMG values were further processed by customized MatLab code (Appendix VIII) and expressed as the percentage of the lunge cycle. The EMG variables were summarized and listed in Table 6.2.

Cross-correlation was also employed to examine the motor activation patterns between the two EMG waveforms over time by sequentially shifting one forward and backward to determine the point of maximum correlation, when the waveforms are most in phase (William, et al., 2013; Wren, et al., 2006). The amount of shift indicates the lag between the two EMG waveforms (Li & Caldwell, 1999; Williams, et al., 2013).

All EMG variables of the 5 lunge attack trails by each fencer were averaged. Means and standard deviation for each fencing group (i.e. Category A and Category B) were further calculated for analysis.

Table 6.3 Summary of EMG dependent variables

EMG variables (unit)	Remark
Amplitude domain • Peak EMG (%IMVC) • Integrated EMG (%IMVC)	Amplitude normalized against IMVC of each tested muscle
Time domain • Onset (% of the lunge cycle) • Occurrence of peak EMG (% of the lunge cycle)	Time-normalized to percentage lunge cycle
Cross-correlation coefficient • R (no unit) • Time lag (% of the lunge cycle)	A total of 28 sets of inter-muscle motor pattern comparison from 8 upper limb muscles were generated for analysis.

6.2.6 Statistical analysis

The Normal Gaussian distribution of all kinematic data was ascertained by the Shapiro-Wilk test. Two-way repeated measures analysis of variance (ANOVA) [2 fencing groups x 4 fencing distances] were used to determine any differences in mean values of the parameters of lunge duration, angular displacement, peak linear velocity, peak angular velocity, peak cross-correlation coefficients and time lag within the 4 fencing distances and between the two fencing groups. For EMG variables, two-way repeated measures analysis of variance (ANOVA) [2 fencing groups x 4 fencing distances] were also employed to examine any differences in the onset of EMG, occurrence of peak EMG, peak EMG, iEMG, peak cross-correlation coefficient and time lag between the two fencing groups. For significant ANOVA results, post-hoc Tukey's honest significant test (for the effect of fencing distance) and paired t-test (for the effect of fencing group) with Bonferroni corrections were followed. The significance level was set at $p<0.05$. SPSS for Window (version 20.0) was used for all statistical tests.

6.3 Results

6.3.1 Kinematic results

6.3.1.1 Lunge duration

Lunge duration of the two fencing groups at various fencing distance were showed in Table 6.4. The lunge duration between fencing groups and amongst distances were different ($F_{3,84}=109.543$; $p<0.001$).The mean lunge duration increased significantly only at longer distances in both fencing groups (Category A: Distance_100 = 0.47±0.16 s, Distance_105 = 0.52±0.11 s, Distance_110 = 0.55±0.08 s and Distance_115 = 0.58±0.09 s; Category B: Distance_100 = 0.44±0.10 s, Distance_105 = 0.54±0.07 s, Distance_110 = 0.60±0.05 s and Distance_115 = 0.63±0.05 s; $F_{1,28}$=74.96; $p<0.001$). Post hoc analysis indicated that there was significant longer lunge duration for Category B fencers as compared to Category A fencers at Distance_110 (p=0.041, CI= 0.003 to 0.110) and Distance_115 (p=0.041; CI: 0.002 to 0.116).

Table 6.4 Lunge duration (second) for Category A and Category B fencers at various fencing distances

	Distance_100	Distance_105	Distance_110	Distance_115	Within-group difference
Category A$^{\alpha}$	0.47±0.16	0.52±0.11	**0.55±0.08**	**0.58±0.09**	†b,c,e,f
Category B$^{\alpha}$	0.44±0.10	0.54±0.07	**0.60±0.05***	**0.63±0.05***	†a,b,c,d,e,f

$^{\alpha}$indicates significant interaction effect (group x fencing distance), p<0.05

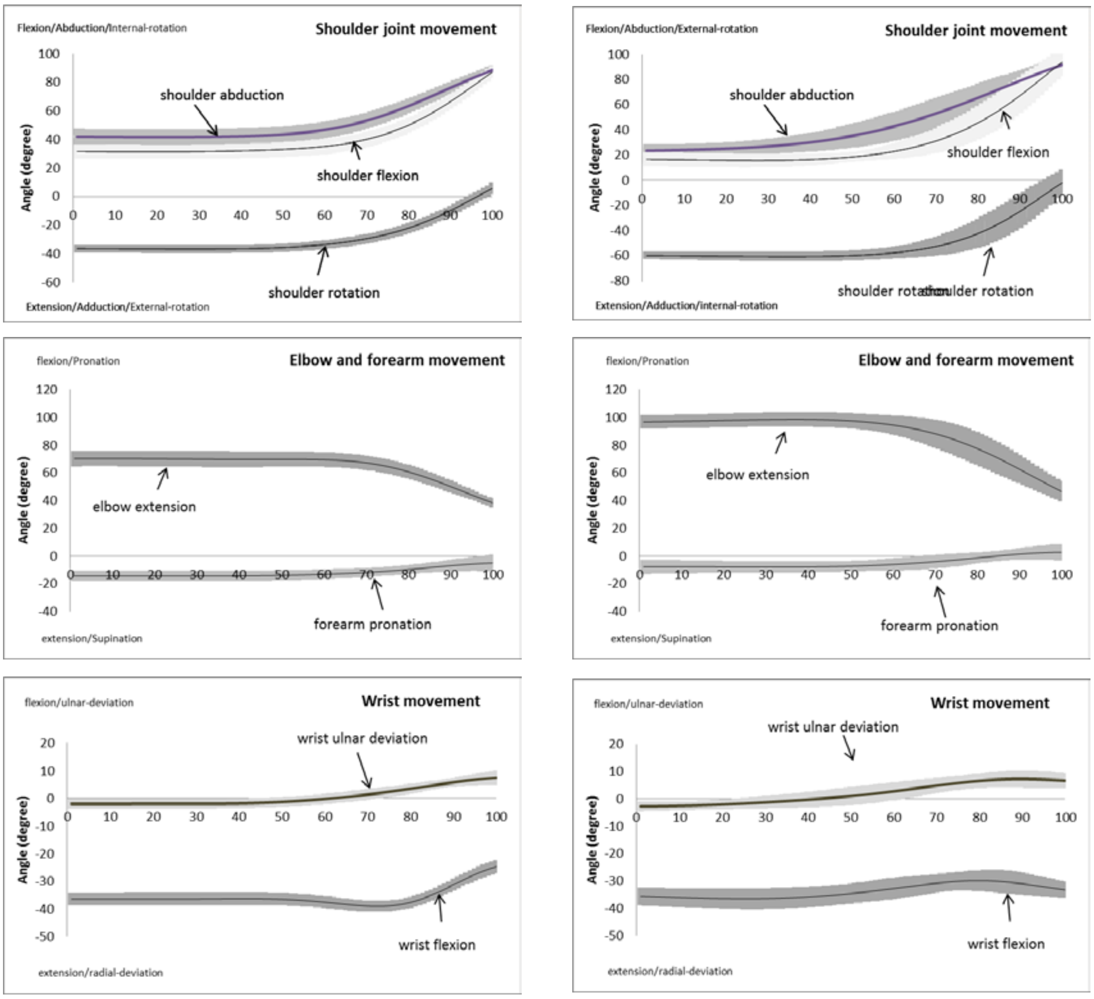
*indicates significant difference (main effect: group) between Category A and Category B fencers, p<0.05

†indicates significant difference (main effect: fencing distance), a: Distance_100 vs Distance_105; b: Distance_100 vs Distance_110; c: Distance_100 vs Distance_115; d: Distance_105 vs Distance_110; e: Distance_105 vs Distance_115; f: Distance_110 vs Distance_115

6.3.1.2 Movement pattern during lunge cycle

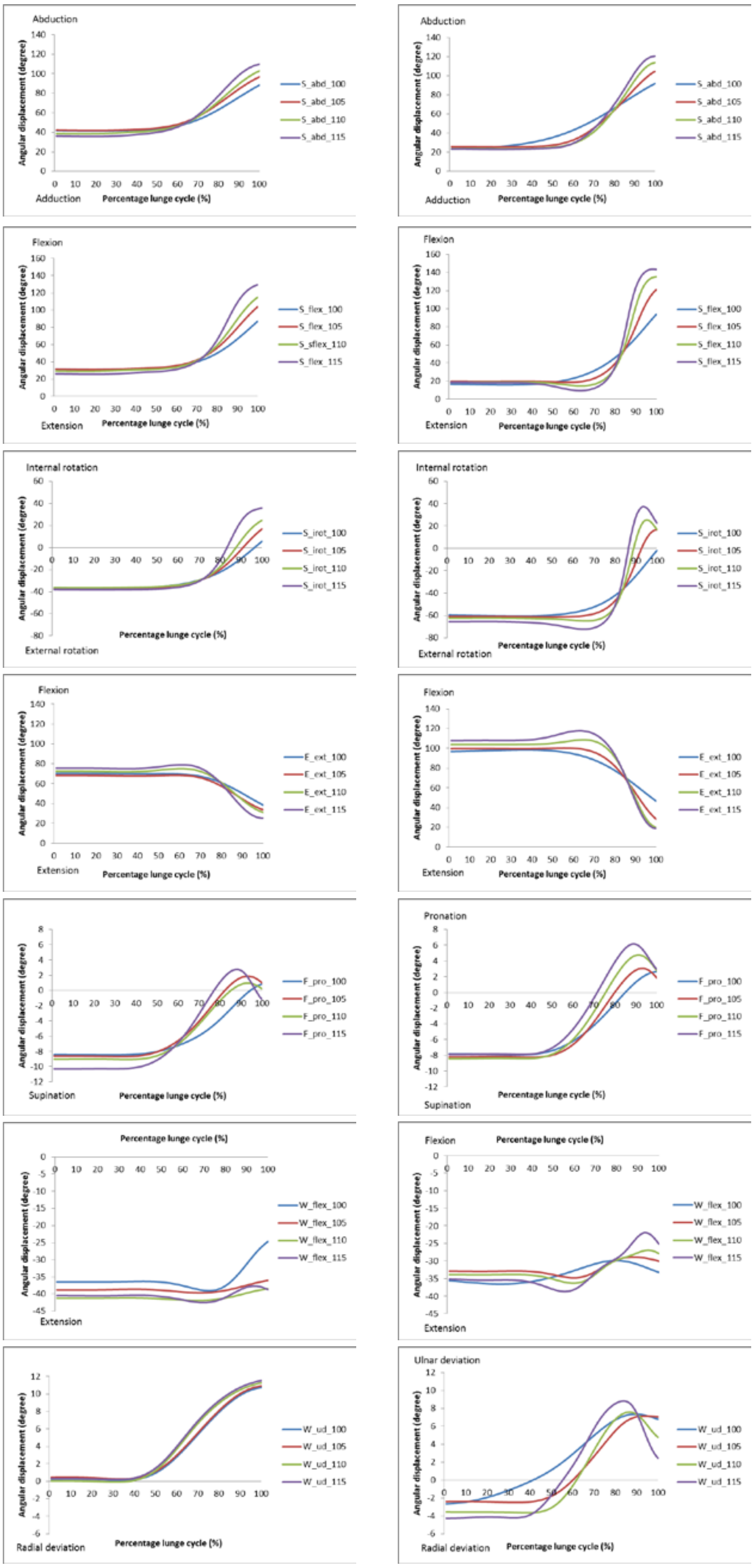
Both fencing groups showed a very similar movement pattern. A typical movement patterns of the two fencing groups were shown in Figure 6.1. At the start of lunge position ("en garde"), fencers kept their shoulders in an abducted (Category A: 31.9°±6.8°, Category B: 28.8°±6.9°), flexed (Category A: 18.3°±9.3°, Category B: 19.7°±7.3°) and externally-rotated (Category A: 45.8°±5.3°, Category B: 42.1°±0.2°)

position. The elbow was flexed (Category A: 78.5°±5.8°, Category B: 82.5°±11.4°) with forearm slightly supinated (Category A: 9.4°±2.4°, Category B: 9.0°±2.4°), wrist extended (Category A: 37.2° ± 2.0°, Category B: 41.0° ± 3.3°) and radial-deviated (Category A: 4.3°±5.5°, Category B: 4.4°±2.6°). From this initial position, fencers started off the lunge by initiating shoulder abduction, flexion and internal rotation, followed by elbow extension to advance the weapon towards the target. When the weapon approached the target, fencers pronated their forearm with wrist flexed and ulnar-deviated to control the foil to hit onto the target (Figure 6.1). Motion patterns for lunge attack were remarkably similar between the two groups at various fencing distances (Figure 6.2).



Category A fencers | Category B fencers

Figure 6.1 Angular displacement-time history curves of the shoulder (top), elbow, forearm (middle) and wrist (bottom) joints at Distance_100. The black lines represent the mean and the grey areas represent the standard deviation



Shoulder abduction

Shoulder flexion

Shoulder internal rotation

Elbow extension

Forearm pronation

Wrist flexion

Wrist ulnar deviation

Category A fencer

Category B fencer

Figure 6.2 Representative time-normalized angular-displacement profiles of a Category A (left) and a Category B (right) fencer at various fencing distances.

6.3.1.3 Angular displacement

Mean angular displacement for the various upper limb joint motions during lunge attack by the two groups of wheelchair fencers were illustrated in Figure 6.1 and 6.2, and summarized in Table 6.5. There were significant interaction effect for shoulder abduction ($F_{3,84}$=9.000; p<0.001), flexion ($F_{3,84}$=16.498; p<0.001) and internal rotation ($F_{3,84}$=18.470; p<0.001). Statistically significant main effects existed between the two fencing groups for shoulder abduction, flexion and rotation; with post-hoc analysis revealed Catgory B fencers have significant larger shoulder flexion (Distance_110: Category A = 99.8±13.9° versus Category B = 114.2±8.8°, p=0.02; Distance_115: Category A = 107.2±9.4° versus Category B = 124.3±8.1°; p<0.001), shoulder abduction (Distance_110: Category A = 74.9±8.1° versus Category B = 84.8±7.3°; p=0.002; Distance_115: Category A = 79.7±6.3° versus Category B = 91.9±5.2°; $F_{1,28}$=34.32; p<0.001) and shoulder rotation (Distance_110: Category A = 60.5±13.8° versus Category B = 82.9±7.7°, p<0.001; Distance_115: Category A = 69.0±19.5° versus Category B = 90.0±15.1°, p=0.003) angular displacements than that of Category A fencers at Distance_110 and Distance_115. There were no significant differences in angular displacement of the elbow motion (Interaction effect: $F_{3,84}$=2.4195; p=0.666, Main-effect for group: $F_{1,28}$=3.499; p=0.063 and Main-effect for fencing distance: $F_{1,28}$=1.0805; p=0.124), forearm motion (Interaction effect: $F_{3,84}$=0.491; p=0.690, Main-effect for group: $F_{1,28}$=0.020; p=0.888 and Main-effect for fencing distance: $F_{1,28}$=0.055; p=0.816) and wrist motion (Interaction effect: $F_{3,84}$=0.877; p=0.456, Main-effect for group: $F_{1,28}$=3.223; p=0.083 and Main-effect for fencing distance: $F_{1,28}$=0.652; p=0.426).

Table 6.5 Upper limb angular displacement (measuring unit: degree) of the Category A and Category B fencers at different fencing distances

	Category A (N=15)				Category B (N=15)				Within-group Differences
	Distance_100	Distance_105	Distance_110	Distance_115	Distance_100	Distance_105	Distance_110	Distance_115	
S_abd$^{\alpha}$	63.7±12.4	72.0±11.7	74.9±8.1	79.7±6.3	61.9±6.2	76.1±5.4	**84.8±7.3***	**91.9±5.2***	†CA: a,b,c,e †CB: a, b, c,d,e,f
S_flex$^{\alpha}$	81.5±18.7	95.9±20.8	99.8±13.9	107.2±9.4	82.2±10.1	100.7±6.5	**114.2±8.8***	**124.3±8.1***	†CA: a,b,c,e,f †CB: a,b,c,d,e,f
S_irot$^{\alpha}$	57.1±17.2	67.1±22.6	60.5±13.8	69.0±19.5	51.9±10.3	75.3±5.5	**82.9±7.7***	**90.0±15.1***	†CA: a, c,f †CB: a,b,c,d,e,f
E_ext	52.0±14.3	59.2±17.5	65.6±13.6	70.0±11.9	47.9±5.8	61.7±10.7	70.7±15.2	72.7±11.4	CA: a, b,c,e,f CB: a,b,c,d,e,f
F_pro	12.1±2.9	15.0±4.7	19.3±7.1	24.6±9.0	10.6±2.2	14.2±3.7	21.7±12.3	23.3±16.5	CA: c,e CB: b,c,d,e
W_flex	9.3±5.4	12.5±10.7	11.0±6.5	14.1±7.7	6.7±2.5	7.1±3.7	9.4±4.0	11.9±5.1	CA: c CB: b,c
W_ud	10.4±2.1	12.3±4.0	11.5±1.5	10.4±1.6	10.1±4.6	10.4±4.0	11.0±4.7	11.7±4.8	CA: - CB: c

S_abd = shoulder abduction, S_flex = shoulder flexion, S_irot = internal shoulder rotation, E_ext = elbow extension, F_pro = forearm pronation, W_flex = wrist flexion, W_ud = wrist ulnar deviation; $^{\alpha}$indicates significant interaction effect (group x fencing distance), $p<0.05$

*indicates significant difference (main effect: group) between Category A and Category B fencers, $p<0.05$

†indicates significant difference (main effect: fencing distance), a: Distance_100 vs Distance_105; b: Distance_100 vs Distance_110; c: Distance_100 vs Distance_115; d: Distance_105 vs Distance_110; e: Distance_105 vs Distance_115; f: Distance_110 vs Distance_115

6.3.1.4 Peak velocity

6.3.1.4.1 Linear horizontal velocity

The peak linear horizontal velocities at the four fencing distances and the two groups were summarized in Table 6.6. Repeated measure ANOVA indicated that there were significant differences between the two fencing groups, lunge distances and the interaction of both groups with distances for the peak horizontal velocity measured at shoulder ($F_{3,84}$=13.991; p=0.000), elbow ($F_{3,84}$=13.826; p=0.000) and wrist ($F_{3,84}$=7.086; p=0.002) joints. Peak linear horizontal velocity for shoulder, elbow and wrist joints significantly increased with lunge distance in both fencing groups (Table 6.6). There was significant difference in peak horizontal linear horizontal velocity between the two fencing groups at Distance_105, Distance_110 and Distance_115. Post-hoc analysis (Table 6.6) revealed that Category B fencers have significantly lower peak horizontal velocity than that of Category A fencers of the shoulder, elbow and wrist at Distance_105 (Shoulder: Category B: 1.10±0.03m/s versus Category A: 1.31±0.04 m/s; Elbow: Category B: 2.14±0.28 m/s versus Category A: 2.62±0.37 m/s; Wrist: Category B: 2.43±0.33 m/s versus Category A: 2.89±0.40 m/s, all p-values<0.001), Distance_110 (Shoulder: Category B: 1.15±0.11 m/s versus Category A: 1.50±0.19 m/s; Elbow: Category B: 2.20±0.40 m/s versus Category A: 2.72±0.27 m/s, Wrist: Category B: 2.54±0.57 m/s versus Category A:

3.03±0.25 m/s, all p-values<0.001) and Distance_115 (Shoulder: Category B: 1.21±0.12m/s versus Category A: 1.65±0.16 m/s, p=0.002; Elbow: Category B: 2.22±0.45 m/s versus Category A: 2.93±0.34 m/s, p=0.005; Wrist: Category B: 2.56±0.67 m/s versus Category A: 3.30±0.35 m/s, p=0.001). Significant within-group differences (all p values <0.05) of the peak linear horizontal velocity were also found in both groups for shoulder, elbow and wrist joints (Table 6.6).

Table 6.6 Peak horizontal velocity (m/s) of shoulder, elbow and wrist joints of the Category A and Category B fencers

	Distance_100	Distance_105	Distance_110	Distance_115	Within-group differences
Shoulder$^{\alpha}$					
Category A	1.12±0.12	**1.31±0.04**	**1.50±0.19**	**1.65±0.16**	†a,b,c,d,e,f
Category B	1.03±0.13	**1.10±0.03***	**1.15±0.11***	**1.21±0.12***	†c,e,f
Elbow$^{\alpha}$					
Category A	2.12±0.31	**2.62±0.37**	**2.72±0.27**	**2.93±0.34**	†a,b,c,e,f
Category B	1.99±0.41	**2.14±0.28***	**2.20±0.40***	**2.22±0.45***	†b,c
Wrist$^{\alpha}$					
Category A	2.41±0.32	**2.89±0.40**	**3.03±0.25**	**3.30±0.35**	†a,b,c,e,f
Category B	2.22±0.41	**2.43±0.33***	**2.54±0.57***	**2.56±0.67***	†a, b, c

$^{\alpha}$indicates significant interaction effect (group x fencing distance), p<0.05

*indicates significant difference (main effect: group) between Category A and Category B fencers, p<0.05

†indicates significant difference (main effect: fencing distance), a: Distance_100 vs Distance_105; b: Distance_100 vs Distance_110; c: Distance_100 vs Distance_115; d: Distance_105 vs Distance_110; e: Distance_105 vs Distance_115; f: Distance_110 vs Distance_115

6.3.1.4.2 Linear vertical velocity

Table 6.7 summarized the peak linear vertical velocities at the four fencing distances and the two groups. There was no significant difference found between group, among distance or interaction of group with distance in regard to the peak linear vertical velocity (Shoulder: $F_{3,84}$=1.176; p=0.34, Elbow: $F_{3,84}$=1.883; p=0.139 and Wrist: $F_{3,84}$=2.877; p=0.540).

Table 6.7 Peak vertical velocity (m/s) of shoulder, elbow and wrist joints of the Category A and Category B fencers

	Distance_100	Distance_105	Distance_110	Distance_115
Shoulder (-ve sign indicates fencers' shoulder joints moved in a downward direction)				
Category A	-0.25±0.06	-0.31±0.11	-0.41±0.06	-0.47±0.02
Category B	-0.25±0.05	-0.35±0.07	-0.38±0.07	-0.44±0.08
Elbow (+ve sign indicates fencers' elbow joints moved in a upward direction)				
Category A	1.51±0.36	1.61±0.55	1.63±0.64	1.49±0.66
Category B	1.52±0.12	1.50±0.18	1.35±0.36	1.22±0.43
Wrist (+ve sign indicates fencers' wrist joints moved in a upward direction)				
Category A	1.25±0.37	1.24±0.37	1.28±0.63	1.02±0.55
Category B	1.36±0.16	1.15±0.20	0.92±0.18	0.79±0.20

6.3.1.4.3 Angular velocity

The peak angular velocities at the four fencing distances and the two groups are summarized in Table 6.8. There was a significant group-by-distance interaction for shoulder flexion ($F_{3,84}$ = 2.221; p=0.042), shoulder abduction ($F_{3,84}$=12.66; p=0.029) and elbow extension ($F_{3,84}$ = 2.154; p= 0.044). The peak angular velocity of the shoulder abduction, shoulder flexion and elbow extension in the two fencing groups significantly increased with the distance, p<0.05. When comparing between the two groups, there was significant lower peak shoulder abduction angular velocity at all fencing distances as compared to Category A fencers (Distance_100: Category B= 393.4 ± 46.7 °/s versus Category A= 471.7 ± 88.4 °/s, p=0.005; Distance_105: Category B = 425.5 ± 52.2 °/s versus Category A= 503.6 ± 98.6 °/s, p=0.011 , Distance_110: Category B= 402.4 ± 94.1 °/s versus Category A= 508.7 ± 87.4 °/s, p=0.003 , Distance_115: Category B=391 ± 114.2 °/s versus Category A= 511.3 ± 82.6 °/s, p= 0.003). For the shoulder flexion, Category B fencers also demonstrated significantly lower peak angular velocities than that of Category A at Distance_110 (Category B= 767.0 ± 287.7 °/s versus Category A= 1016.2 ± 189.2 °/s, p=0.009) and Distance_115 (Category B = 782.0 ± 306.6 °/s versus Category A = 1065.7 ± 108.4 °/s, p=0.002). The peak angular velocity of the elbow extension was also shown to be significantly lower in Category B fencers at Distance_110 (Category B= 466.4 ± 129.4 °/s versus Category A= 569.3 ± 105.2 °/s, p=0.024) and Distance_115 (Category B = 476.2 ± 99.4 °/s versus Category A = 598.2 ± 94.4 °/s, p=0.002).

Table 6.8 Peak angular velocity (degree/s) of the Category A and Category B fencers at different fencing distances

	Distance_100	Distance_105	Distance_110	Distance_115	Within-group differences
Shoulder abduction (+ve) / adduction (-ve) $^{\alpha}$					
Category A	**471.7 ± 88.4**	**503.6 ± 98.6**	**508.7 ± 87.4**	**511.3 ± 82.6**	†a,b,c
Category B	**393.4 ± 46.7***	**425.5 ± 52.2***	**402.4 ± 94.1***	**391 ± 114.2***	†a
Shoulder flexion (+ve) / extension (-ve) $^{\alpha}$					
Category A	797.4 ± 162.2	885.9 ± 293.9	**1016.2 ± 189.2**	**1065.7 ± 108.4**	†b,c
Category B	676.5 ± 192.0	765.4 ± 163.3	**767.0 ± 287.7***	**782.0 ± 306.6***	†c
Shoulder internal rotation (+ve) / external rotation (-ve)					
Category A	638.2 ± 256.4	776.8 ± 296.2	767.9 ± 244.0	800.6 ± 205.1	a
Category B	572.7 ± 185.6	742.7 ± 155.3	801.4 ± 381.4	873.1 ± 526.9	a,b,c
Elbow flexion (+ve) / extension (-ve) $^{\alpha}$					
Category A	-438.8 ± 86.9	-527.0 ± 153.8	**-569.3 ± 105.2**	**-598.2 ± 94.4**	†a,b,c,e
Category B	-382.4 ± 95.8	-460.1 ± 111.6	**-466.4 ± 129.4***	**-476.2 ± 99.4***	†a,b,c
Forearm pronation (+ve) / supination (-ve)					
Category A	74.2 ± 39.7	76.7 ± 40.6	69.6 ± 49.1	69.1 ± 46.1	-
Category B	64.4 ± 32.0	56.9 ± 24.1	62.4 ± 39.9	60.4 ± 18.2	-
Wrist flexion (+ve) / extension (-ve)					
Category A	93.6 ± 57.1	105.8 ± 57.7	106.1 ± 63.5	130.3 ± 48.2	c
Category B	70.3 ± 27.7	93.2 ± 29.3	100.1 ± 37.2	124.2 ± 48.1	c
Wrist ulnar deviation (+ve) /radial deviation(-ve)					
Category A	69.3 ± 22.1	54.0 ± 19.6	57.1 ± 25.8	59.2 ± 56.9	-
Category B	67.2 ± 21.1	64.5 ± 20.9	57.9 ± 19.1	56.0 ± 20.4	-

$^{\alpha}$indicates significant interaction effect (group x fencing distance), $p<0.05$

*indicates significant difference (main effect: group) between Category A and Category B fencers, $p<0.05$

†indicates significant difference (main effect: fencing distance), a: Distance_100 vs Distance_105; b: Distance_100 vs Distance_110; c: Distance_100 vs Distance_115; d: Distance_105 vs Distance_110; e: Distance_105 vs Distance_115; f: Distance_110 vs Distance_115

6.3.1.5 Inter-joint coordination

Only the results with inter-joint cross-correlation coefficient values over 0.8 were listed and summarized in Table 6.9. High cross-correlation coefficients (R) were found in shoulder flexion-elbow extension (ranged from 0.87-0.91 for Category A and 0.79-0.92 for Category B), shoulder abduction-elbow extension (ranged from 0.84-0.87 for Category A and 0.74-0.81 for Category B) and elbow extension-forearm pronation (ranged from 0.95-0.98 for Category A and 0.94-0.98 for Category B) for both fencing groups, indicating the highly coordination of these inter-joint movements during the fencing lunge attack.

The cross-correlation values showed significantly lower for shoulder flexion-elbow extension and shoulder-elbow extension in Category B fencers at the longer fencing distance of Distance_110 and Distance_115.

Also, there were significant increases in mean delay (lags) between shoulder flexion and elbow extension as well as shoulder abduction and elbow extension in Category B fencers at longer fencing distances of Distance_110 and Distance_115 as compared to Category A fencers, $p<0.05$.

6.3.1.6 Overall kinematic results

All kinematic results were summarized in Table 6.10 and Table 6.11. Table 6.10 illustrated the between-group difference for Category A and B fencers while Table 6.11 depicited the within-group differences in Category A and B fencers at different fencing distances.

Table 6.9 Mean inter-joint cross-correlation coefficients (R) and time lag (% lunge cycle) of the Category A and Category B fencers at different fencing distances

	Distance_100		Distance_105		Distance_110		Distance_115		Within-group difference	
	R	Lag (%)	R	Lag (%)	R	Lag (%)	R	Lag (%)	R	Lag (%)
Shoulder flexion and elbow extension$^{\alpha}$										
Category A	-0.91±0.04	8.13±2.90	-0.88±0.02	7.80±4.02	**-0.87±0.06**	**12.53±3.29**	**-0.88±0.09**	**16.07±2.76**	-	†b,c,d,e,f
Category B	-0.92±0.06	8.40±4.42	-0.89±0.08	10.73±3.99	**-0.83±0.11***	**17.87±2.13***	**-0.79±0.14***	**20.20±3.34***	†b,c,d,e,f	†b,c,d,e,f
Shoulder abduction and elbow extension$^{\alpha}$										
Category A	-0.86±0.06	9.93±2.12	-0.84±0.07	12.20±3.32	**-0.87±0.05**	16.10±2.67	**-0.86±0.11**	**19.90±3.42**	-	†b,c,d,e,f
Category B	-0.81±0.11	9.87±5.34	-0.83±0.06	11.73±1.54	**-0.76±0.11***	18.67±4.10	**-0.74±0.16***	**24.09±2.63***	†b,c,d,e,f	†b,c,d,e,f
Elbow extension and forearm pronation										
Category A	0.97±0.02	0.00±0.00	0.98±0.03	0.01±0.00	0.96±0.02	0.00±0.00	0.95±0.04	0.00±0.01	-	-
Category B	0.98±0.16	0.01±0.00	0.97±0.09	0.00±0.00	0.95±0.05	0.01±0.00	0.94±0.06	0.00±0.00	-	-

+ve: movements in-phase; -ve: movements out-of-phase

$^{\alpha}$indicates significant interaction effect (group x fencing distance), $p<0.05$

*indicates significant difference (main effect: group) between Category A and Category B fencers, $p<0.05$

†indicates significant difference (main effect: fencing distance), a: Distance_100 vs Distance_105; b: Distance_100 vs Distance_110; c: Distance_100 vs Distance_115; d: Distance_105 vs Distance_110; e: Distance_105 vs Distance_115; f: Distance_110 vs Distance_115

Table 6.10 Summary of the between-group differences of the kinematic variables in Category A and Category B fencers at various fencing distances

Kinematic variable (unit)		Category A vs Category B			
		Distance_100	Distance_105	Distance_110	Distance_115
Lunge duration (second)		-	-	**CB ↑***	**CB ↑***
Angular displacement (degree)	S_abd	-	-	**CB ↑***	**CB ↑***
	S_flex	-	-	**CB ↑***	**CB ↑***
	S_irot	-	-	**CB ↑***	**CB ↑***
	E_ext	-	-	-	-
	F_pro	-	-	-	-
	W_flex	-	-	-	-
	W_ud	-	-	-	-
Peak velocity (horizontal/ vertical) (m/s)	Shoulder	-	**CB ↓***/ -	**CB ↓***/ -	**CB ↓***/ -
	Elbow	-	**CB ↓***/ -	**CB ↓***/ -	**CB ↓***/ -
	Wrist	-	**CB ↓***/ -	**CB ↓***/ -	**CB ↓***/ -
Angular velocity (degree/s)	S_abd	**CB ↓ ***	**CB ↓***	**CB ↓***	**CB ↓***
	S_flex	-	-	**CB ↓***	**CB ↓***
	S_irot	-	-	-	-
	E_ext	-	-	**CB ↓***	**CB ↓***
	F_pro	-	-	-	-
	W_flex	-	-	-	-
	W_ud	-	-	-	-
Inter-joint cross-correlation (R/lag)	S_flex Vs E_ext	- / -	- / -	**CB ↓*** / **CB ↑***	**CB ↓*** / **CB ↑***
	S_abd Vs E_ext	- / -	- / -	**CB ↓*** / -	**CB ↓*** / **CB ↑***
	E_ext Vs F_pro	- / -	- / -	- / -	- / -

*p<0.05

Table 6.11 Summary of the within-group differences of the kinematic variables in Category A and Category B fencers at various fencing distances

Kinematic variable (unit)		Distance_100 vs Distance_105		Distance_100 vs Distance_110		Distance_100 vs Distance_115		Distance_105 vs Distance_110		Distance_105 vs Distance_115		Distance_110 vs Distance_115	
		CA	CB	CA	CB	CA	CB	CA	CB	CA	CB	CA	CB
Lunge duration (second)		-	*	*	*	*	*	-	*	*	*	*	*
Angular displacement (degree)	S_abd	*	*	*	*	*	*	-	*	*	*	-	*
	S_flex	*	*	*	*	*	*	-	*	*	*	*	*
	S_irot	*	*	-	*	*	*	-	*	-	*	*	*
	E_ext	*	*	*	*	*	*	-	*	*	*	*	*
	F_pro	-	-	-	*	*	*	-	*	*	*	-	-
	W_flex	-	-	-	*	*	*	-	-	-	-	-	-
	W_ud	-	-	-	-	-	*	-	-	-	-	-	-
Peak velocity (horizontal) (m/s)	Shoulder	*	-	*	-	*	*	*	-	*	*	*	*
	Elbow	*	-	*	*	*	*	-	-	*	-	*	-
	Wrist	*	*	*	*	*	*	-	-	*	-	*	-
Angular velocity (degree/s)	S_abd	*	*	*	-	*	-	-	-	-	-	-	-
	S_flex	-	-	*	*	*	-	-	-	-	-	-	-
	S_irot	*	*	-	*	-	*	-	-	-	-	-	-
	E_ext	*	*	*	*	*	*	-	-	*	-	-	-
	F_pro	-	-	-	-	-	-	-	-	-	-	-	-
	W_flex	-	-	-	-	*	*	-	-	-	-	-	-
	W_ud	-	-	-	-	-	-	-	-	-	-	-	-
Inter-joint cross-correlation (R/lag)	S_flex vs E_ext	-/-	-/-	-/ *	*/ *	-/ *	*/ *	-/ *	*/ *	-/ *	*/ *	-/ *	*/ *
	S_abd Vs E_ext	-/-	-/-	-/ *	*/ *	-/ *	*/ *	-/ *	*/ *	-/ *	*/ *	-/ *	*/ *
	E_ext Vs F_pro	-/-	-/-	-/-	-/-	-/-	-/-	-/-	-/-	-/-	-/-	-/-	-/-

*p<0.05

6.3.2 EMG results

6.3.2.1 Fencing arm muscle activation pattern

Figure 6.3 illustrated the typical SEMG patterns of the 8 fencing arm muscles against the displacement-time curve of the corresponding upper limb joints of a Category A fencer during lunge at Distance_100. Qualitative analysis showed that UT, INF, ANT, MID and BIC activated first for initiating the shoulder flexion, abduction and internal-rotation. Then TRI extended the elbow. As the foil approaches the target, WF and WE activated almost simultaneously to control the weapon for target hitting. Contraction of UT, INF, DEL, BIC and TRI remained active throughout the entire lunge cycle after the initial activation. Similar activation sequence was denoted in both Category A and B fencers at longer fencing distances (Figure 6.4).

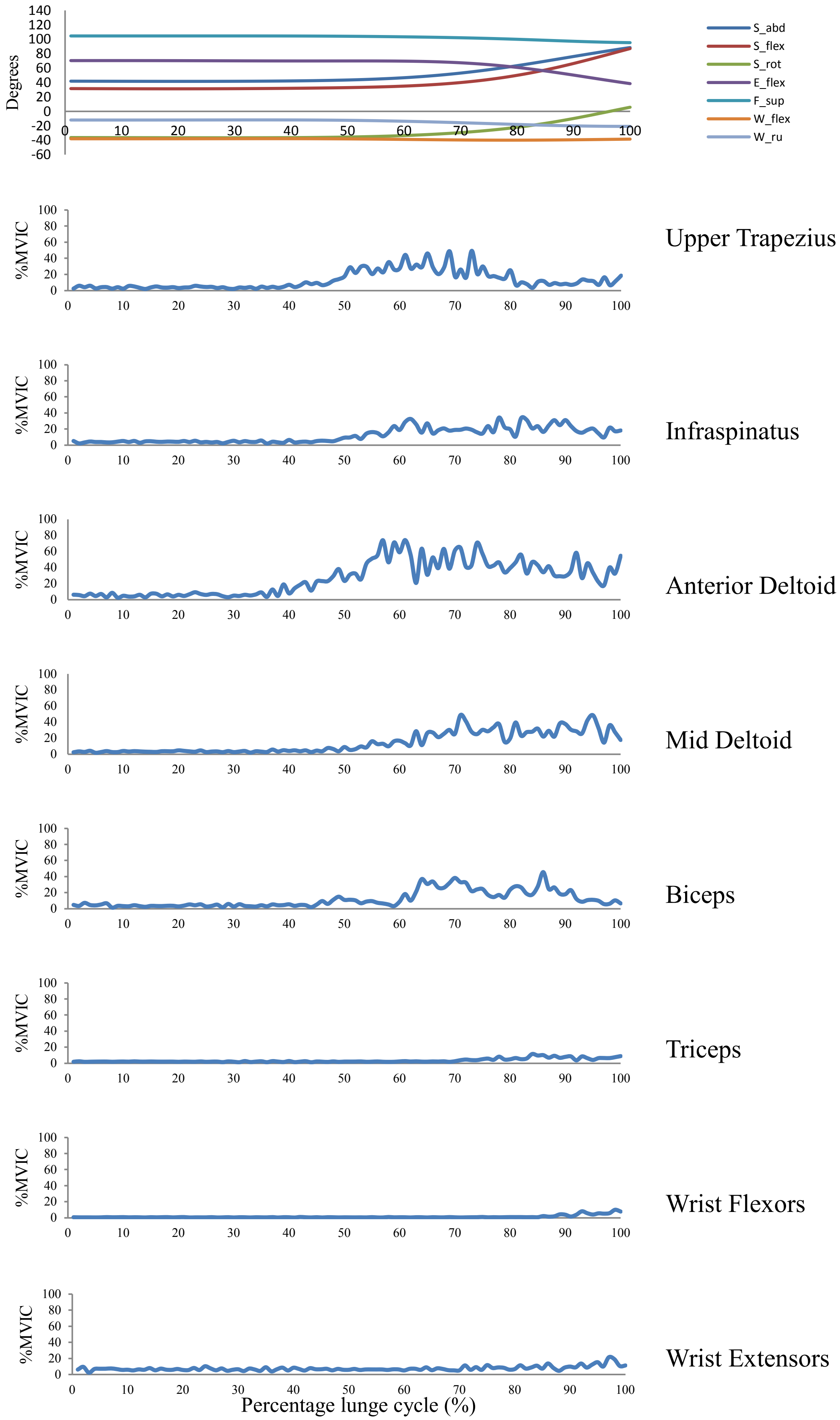
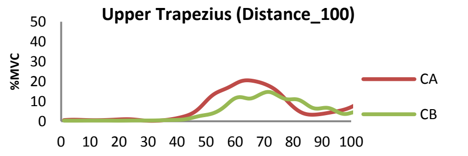
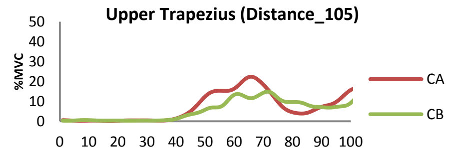
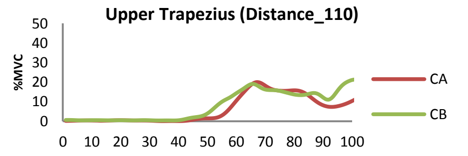


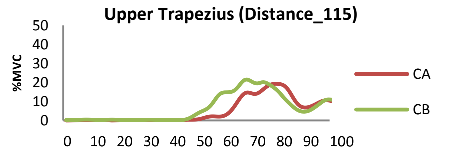
Figure 6.3 An illustrative EMG waveforms of a Category A fencer who lunge at Distance_100

Upper Trapezius

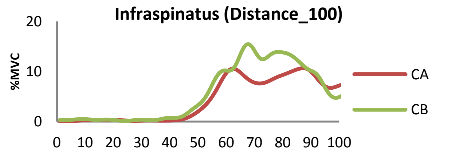


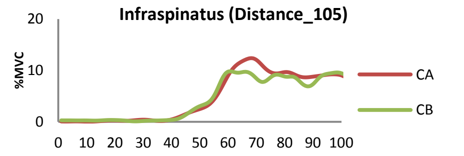


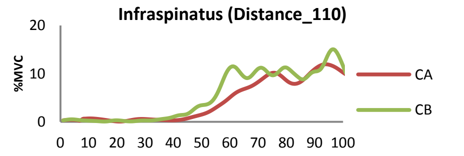


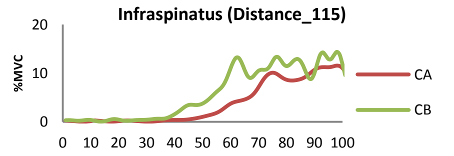


Infraspinatus

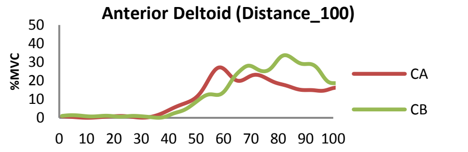


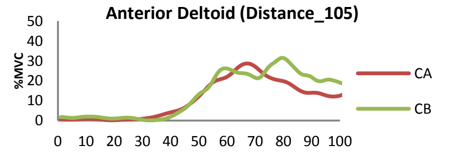


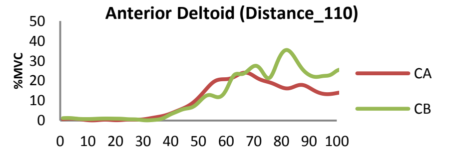


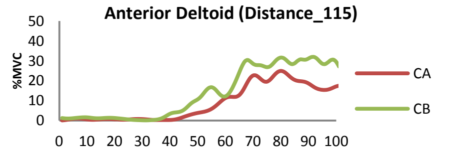


Anterior-Deltoid

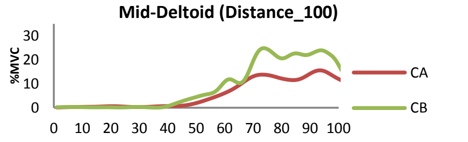


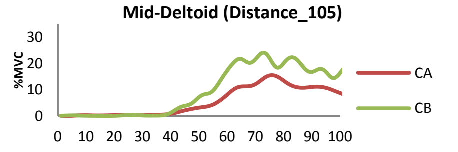


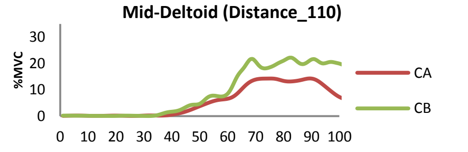


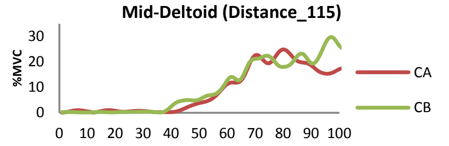


Mid-Deltoid









Distance_100	Distance_105	Distance_110	Distance_115

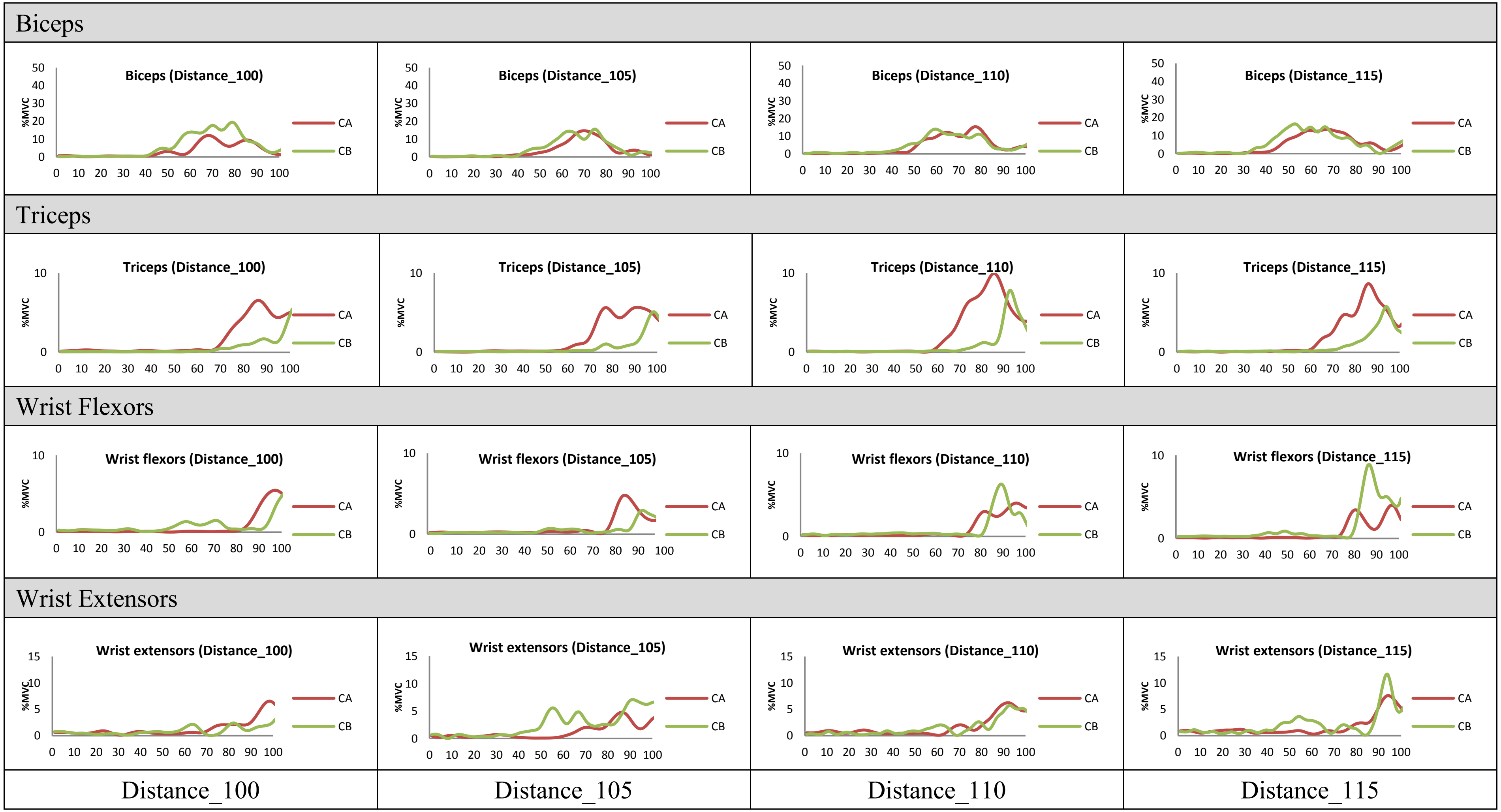


Figure 6.4 Typical EMG waveforms of the 8 tested fencing arm muscles (amplitude normalized by MVIC and expressed as %MVIC and time-normalized to % lunge cycle) as recorded in a Category A and B fencer who execute lunge attack at Distance_100, Distance_105, Distance_110 and Distance_115. CA: Category A, CB: Category B fencers

6.3.2.2 Onset and occurrence of peak EMG

Quantitative evaluation for the mean onset of EMG signals (as expressed in % lunge cycle) showed a proximal-to-distal motor recruitment sequence from shoulder muscles (UT, DEL) and BIC to TRI and then to the WF and WE during the lunge attack (Table 6.12). As the fencing distance increased, all muscles continued to follow the proximal-to-distal recruitment patterns, although they all adopted an earlier motor recruitment. In general, muscles of the Category B fencers recruited earlier when they lunge at longer distances (Table 6.13). There was significant group by distance interaction effect for INF ($F_{3,84}$=4.435, p=0.011), MID ($F_{3,84}$=5.483, p=0.006), BIC ($F_{3,84}$=4.413, p=0.009), TRI ($F_{3,84}$=8.495, p=0.001), WF ($F_{3,84}$=3.635, p=0.033) and WE ($F_{3,84}$=13.424, p<0.001) on the mean onset EMG. Compared to the Category A fencers, the Category B fencers showed significantly earlier recruitment of the INF, MID, BIC, TRI, WF and WE at Distance_110 and Distance_115 (p<0.05). Specifically, the post-hoc tests showed that the mean EMG onset of INF, MID, BIC, TRI, WF and WE in Category B fencers ranged from a lower end of 2.0 to a higher end of 11.6% earlier than those of the Category A fencers at Distance_110 and Distance_115 (Table 6.13).

The occurrence of peak EMG values also demonstrated a similar motor pattern with proximal muscles reached the peak EMG values first before the distal muscles (Table 6.14). There was significant group by distance interaction effect for BIC

($F_{3,84}$=26.403, p<0.001), TRI ($F_{3,84}$=31.943, p<0.001), WF ($F_{3,84}$=18.917, p<0.001) and WE ($F_{3,84}$=30.338, p<0.001) for the occurrence of peak EMG values. Interestingly, muscles over the upper arm and forearm (BIC, TRI, WF and WE) showed a significantly earlier occurrence of peak EMG values at Distance_110 and Distance_115 (Table 6.14). Explicitly, BIC reached the peak EMG levels even as early as to the fencing distances of Distance_105, Distance_110 and Distance_115. Post-hoc tests between the two fencing groups showed that the mean peak EMG onset of BIC, TRI, WF and WE in Category B fencers ranged from 12.1% to 19.2% earlier than those of the Category A fencers at Distance_110 and Distance_115 (Table 6.15).

Table 6.12 Mean EMG onset (% lunge cycle) for Category A and Category B fencers at different fencing distances

Peak at	Category A (n=15)				Category B (n=15)				Within-group differences
	Distance_100	Distance_105	Distance_110	Distance_115	Distance_100	Distance_105	Distance_110	Distance_115	
Upper Trapezius	38.2±1.8	38.1±3.2	35.5±2.2	36.1±4.4	38.8±2.5	38.2±3.8	34.2±2.8	31.7±5.9	CA[†]: b CB[†]: b,c,d,e
Infraspinatus[α]	41.6±3.7	36.7±2.3	34.8±1.2	35.1±1.5	39.9±2.5	36.0±2.0	**32.7±2.3***	**30.2±1.5***	CA[†]: a,b,c CB[†]: a,b,c,d,e,f
Anterior Deltoid	38.3±2.3	34.3±1.8	34.8±3.1	33.5±2.0	39.2±2.3	35.2±4.9	**32.2±2.9***	31.9±2.7	CA[†]: a,b,c CB[†]: a,b,c,d,e
Mid Deltoid[α]	39.0±4.5	36.3±1.9	34.4±3.1	35.1±3.9	38.8±4.0	**38.0±1.2***	**32.3±2.1***	**30.4±1.6***	CA[†]: b CB[†]: b,c,d,e
Biceps[α]	43.5±4.5	36.1±5.0	36.9±6.9	34.2±3.4	41.6±1.5	38.4±2.2	**32.1±2.8***	**29.6±1.7***	CA[†]: a,b,c CB[†]: b,c,d,e
Triceps[α]	61.3±12.2	57.5±6.1	54.6±4.5	57.2±2.0	65.1±4.7	61.8±3.6	**49.8±2.7***	**48.3±5.5***	CA[†]: b CB[†]: b,c,d,e
Wrist Flexors[α]	75.1±3.4	71.6±2.6	71.1±8.1	71.4±2.5	76.8±5.5	70.4±12.3	**62.5±7.0***	**63.2±4.7***	CA: - CB[†]: b,c
Wrist Extensors[α]	72.2±4.7	70.0±4.1	70.5±6.9	72.8±9.3	73.8±3.5	68.6±6.5	**60.7±6.2***	**61.2±5.3***	CA: - CB[†]: a,b,c,d,e

[α]indicates significant interaction effect (group x fencing distance), $p<0.05$

*indicates significant difference (main effect: group) between Category A and Category B fencers, $p<0.05$

[†]indicates significant difference (main effect: fencing distance), a: Distance_100 vs Distance_105; b: Distance_100 vs Distance_110; c: Distance_100 vs Distance_115; d: Distance_105 vs Distance_110; e: Distance_105 vs Distance_115; f: Distance_110 vs Distance_115

Table 6.13 Difference in EMG onset of the different tested fencing arm muscles between the Category A and the Category B fencers at Distance_110 and Distance_115

Muscle	Difference in EMG onset (% lunge cycle) between the Category A and the Category B fencers at different distance					
	Distance_110	Confidence Interval	p-value	Distance_115	Confidence Interval	p-value
Infraspinatus	2.1	0.7 to 3.4	0.004	4.9	3.8 to 6.1	< 0.001
Mid Deltoid	2.0	0.5 to 4.0	0.045	4.8	2.5 to 7.0	< 0.001
Biceps	4.8	0.8 to 8.7	0.019	4.5	2.5 to 6.6	< 0.001
Triceps	4.8	2.0 to 7.5	0.002	8.8	5.7 to 11.9	< 0.001
Wrist flexors	8.6	3.0 to 14.3	0.004	8.2	5.4 to 11.1	< 0.001
Wrist extensors	9.8	4.9 to 14.7	< 0.001	11.6	6.0 to 17.3	< 0.001

Table 6.14 Mean occurrence of peak EMG value (% lunge cycle) for Category A and Category B fencers at different fencing distances

Peak at	Category A (n=15)				Category B (n=15)				Within-group differences
	Distance_100	Distance_105	Distance_110	Distance_115	Distance_100	Distance_105	Distance_110	Distance_115	
Upper Trapezius	67.7±4.9	64.9±2.1	65.2±14.3	67.4±21.0	70.7±5.8	65.8±6.2	62.6±7.6	67.9±5.3	CA: - CB†: a
Infraspinatus	69.1±14.3	65.7±8.3	68.0±11.7	66.4±6.1	77.5±11.6	62.6±3.4	63.3±9.5	71.2±9.5	CA: - CB†: a,b,f
Anterior Deltoid	73.0±7.9	67.5±6.9	70.1±2.3	69.3±11.5	74.5±9.5	66.4±7.5	67.5±8.7	73.0±8.7	CA: - CB†: e
Mid Deltoid	72.8±2.1	68.3±4.8	73.3±13.0	72.7±10.6	74.7±7.0	73.8±11.7	72.1±7.3	73.0±9.9	CA: - CB: -
Biceps$^{\alpha}$	69.1±2.6	68.9±3.6	66.4±4.1	71.5±2.0	71.6±8.6	**59.4±1.9***	**54.3±6.8***	**57.0±8.3***	CA†: f CB†: a,b,c,d
Triceps$^{\alpha}$	85.5±3.4	80.0±6.1	86.3±3.1	84.7±4.2	**93.2±1.8***	78.6±3.4	**71.2±9.3***	**70.0±6.5***	CA†: a,d CB†: a,b,c,d,e
Wrist Flexors$^{\alpha}$	94.0±2.4	87.3±4.8	90.8±6.5	87.3±8.4	92.7±7.6	90.0±4.8	**78.3±4.7***	**68.9±9.2***	CA†: a,b,c CB†: a,b,c,d,e
Wrist Extensors$^{\alpha}$	93.9±5.8	89.1±5.5	90.8±2.8	88.3±4.1	95.3±3.6	86.2±7.3	**72.5±5.2***	**69.2±5.6***	CA†: a,b,c CB†: a,b,c,d,e

$^{\alpha}$indicates significant interaction effect (group x fencing distance), p<0.05

*indicates significant difference (main effect: group) between Category A and Category B fencers, p<0.05

†indicates significant difference (main effect: fencing distance), a: Distance_100 vs Distance_105; b: Distance_100 vs Distance_110; c: Distance_100 vs Distance_115; d: Distance_105 vs Distance_110; e: Distance_105 vs Distance_115; f: Distance_110 vs Distance_115

Table 6.15 Difference in peak EMG onset of the different tested fencing arm muscles between the Category A and the Category B fencers at Distance_110 and Distance_115

Muscle	Difference in EMG onset (% lunge cycle) between the Category A and the Category B fencers at different distance					
	Distance_110	Confidence Interval	p-value	Distance_115	Confidence Interval	p-value
Biceps	12.1	7.9 to 16.3	<0.001	14.5	10.1 to 19.0	<0.001
Triceps	15.1	9.9 to 20.3	<0.001	14.7	10.6 to 18.8	<0.001
Wrist flexors	12.5	8.2 to 16.7	<0.001	18.4	11.8 to 25.0	<0.001
Wrist extensors	18.3	15.2 to 21.4	<0.001	19.2	15.5 to 22.8	<0.001

6.3.2.3 Peak EMG amplitude and integrated EMG

Peak EMG amplitude for all the tested fencing arm muscles increased in both Category A and B fencers with the fencing distance. The peak EMG values were generally higher for the proximal shoulder muscles including UT, INF, ANT and MID (Category A: ranged from 14.1 to 28.6 %MVIC and Category B: ranged from 14.5% to 39.7%) than the upper arm muscles of BIC and TRI (Category A: ranged from 9.1 to 18.9 %MVIC and Category B: ranged from 15.8 to 29.4 %MVIC). The forearm muscles were found to have generally lower SEMG activity (Category A: 5.9 to 12.1 %MVIC and Category B: 5.2 to 14.3% MVIC). Mean peak EMG values during lunge attack at different fencing distances are summarized in Table 6.16. There was significant group-distance interaction effect for UT ($F_{3,84}$=8.684, p<0.001), INF ($F_{3,84}$=12.525, p<0.001), ANT ($F_{3,84}$=4.120, p=0.030) and MID ($F_{3,84}$=20.444, p<0.001) on the peak EMG values. There were significantly higher peak EMG values of the UT and MID in Category B fencers when they executed the lunge at Distance_110 and Distance_115 than that of Category A. The EMG values of INF and ANT in Category B at Distance_105, Distance_110 and Distance_115 were significantly higher than that of Category A (Table 6.17).

The integrated-EMG values during lunge attack were relatively higher in the proximal shoulder muscles of UT, INF, ANT and MID (Category A: ranged from 23.4 to 49.6 %

MVIC; Category B: ranged from 22.1 to 67.6 %MVIC) as compared to the upper arm and forearm muscles of BIC, TRI, WF and WE (Category A: ranged from 12.7 to 23.8 %MVIC; Category B: ranged from 12.2 to 27.4 % MVIC) (Table 6.18). Statistically significant interaction effects of group by distance were found in UT ($F_{3,84}$=19.060, p<0.001), INF ($F_{3,84}$=27.852, p<0.001), ANT ($F_{3,84}$=26.983, p<0.001) and MID ($F_{3,84}$=78.390, p<0.001). Specifically, iEMG values for UT, INF, ANT and MID in Category B fencers increased significantly as the lunge distance increased (p<0.05). However, this trend was not found in Category A fencers (p values ranged from 0.468-1.000). Category B fencers have significantly higher iEMG values than Category A fencers for INF and MID at Distance_105, Distance_110 and Distance_115, and UT and ANT at Distance_110 and Distance_115 (Table 6.19).

Table 6.16 Peak EMG value (% MVIC) for Category A and Category B fencers at different fencing distances

Peak EMG	Category A (n=15)				Category B (n=15)				Within-group differences
	Distance_100	Distance_105	Distance_110	Distance_115	Distance_100	Distance_105	Distance_110	Distance_115	
Upper Trapezius$^{\alpha}$	14.9±2.9	19.4±2.9	20.9±1.3	20.5±5.3	14.5±2.3	17.9±2.6	**23.9±1.7***	**30.3±8.6***	CA†: a,b CB†: a,b,c,d,e,f
Infraspinatus$^{\alpha}$	14.1±1.9	20.0±1.3	23.4±5.5	28.6±6.1	13.5±1.4	**16.0±1.9***	**31.9±4.5***	**34.2±6.0***	CA†: a,b,c,d,e,f CB†: a,b,c,d,e
Anterior Deltoid$^{\alpha}$	22.0±4.8	24.0±5.3	26.7±5.9	26.1±3.3	25.4±10.8	**30.4±5.3***	**39.7±1.7***	**39.5±2.9***	CA: - CB†: b,c,d,e
Mid-Deltoid$^{\alpha}$	18.8±2.8	19.8±4.8	18.1±1.9	18.0±2.5	21.8±3.2	17.0±1.0	**26.1±5.4***	**29.4±2.8***	CA: - CB†: a,b,c,d,e
Biceps	14.5±3.8	14.5±6.2	18.9±6.2	13.9±4.7	17.0±3.2	15.8±5.0	16.7±4.8	16.8±3.3	CA†: f CB: -
Triceps	9.1±1.5	8.6±1.2	12±2.2	11.5±3.2	9.5±1.6	8.5±3.9	10.8±2.3	11.0±2.8	CA†: b,d CB†: d
Wrist Flexors	7.6±3.0	5.9±3.9	6.7±1.8	7.2±1.7	7.7±3.6	5.2±1.2	7.2±3.9	9.9±3.7	CA: - CB†: e,f
Wrist Extensors	6.7±2.4	9.9±1.5	12.1±2.8	12.0±2.5	5.2±3.4	11.1±1.8	10.3±1.6	14.3±4.4	CA†: a,b CB†: a,c,e

$^{\alpha}$indicates significant interaction effect (group x fencing distance), $p<0.05$

*indicates significant difference (main effect: group) between Category A and Category B fencers, $p<0.05$

†indicates significant difference (main effect: fencing distance), a: Distance_100 vs Distance_105; b: Distance_100 vs Distance_110; c: Distance_100 vs Distance_115; d: Distance_105 vs Distance_110; e: Distance_105 vs Distance_115; f: Distance_110 vs Distance_115

Table 6.17 Difference in Peak EMG values (% MVIC) between the Category A and the Category B fencers at Distance_105, Distance_110 and Distance_115

Muscle	Difference in Peak EMG between the Category A and the Category B fencers at different distance								
	Distance_105	Confidence Interval	p-value	Distance_110	Confidence Interval	p-value	Distance_115	Confidence Interval	p-value
Upper Trapezius	-	-	-	3.1	1.8 to 4.4	<0.001	9.7	3.5 to 15.9	0.004
Infraspinatus	3.6	2.2 to 5.1	<0.001	8.2	3.9 to 12.5	0.001	5.6	4.1 to 10.8	0.036
Anterior Deltoid	6.4	1.8 to 11.0	0.009	13.0	9.2 to 16.7	<0.001	13.4	10.8 to 16.1	<0.001
Mid Deltoid	-	-	-	8.0	4.5 to 11.4	<0.001	11.4	9.2 to 13.7	<0.001

Table 6.18 Integrated EMG value (% MVIC) for Category A and Category B fencers at different fencing distances

Integrated EMG	Category A (n=15)				Category B (n=15)				Within-group differences
	Distance_100	Distance_105	Distance_110	Distance_115	Distance_100	Distance_105	Distance_110	Distance_115	
Upper Trapezius$^{\alpha}$	36.3±4.2	37.7±2.9	38.3±6.6	39.0±4.2	34.3±2.8	39.5±8.7	**47.5±6.3***	**54.1±4.8***	CA: - CB†: b,c,d,e,f
Infraspinatus$^{\alpha}$	24.3±2.2	23.8±0.8	23.4±2.5	25.8±4.3	22.1±2.3	**25.9±5.3***	**32.2±5.4***	**37.8±4.8***	CA: - CB†: a,b,c,d,e,f
Anterior Deltoid$^{\alpha}$	46.9±3.4	46.7±5.9	49.6±5.7	46.7±2.9	43.2±7.0	50.9±8.5	**59.7±4.8***	**67.6±8.6***	CA: - CB†: a,b,c,d,e,f
Mid Deltoid$^{\alpha}$	27.4±1.2	27.2±1.9	26.5±2.4	25.8±3.7	26.9±5.1	**30.8±6.0***	**38.0±3.4***	**46.5±4.8***	CA: - CB†: a,b,c,d,e,f
Biceps	21.8±3.9	23.8±4.0	23.1±4.0	18.5±3.7	20.8±1.6	27.4±7.5	25.9±6.6	**24.4±4.0***	CA†: c CB†: a,,b,c
Triceps	13.3±0.9	12.7±0.7	14.3±1.8	14.6±1.3	12.2±1.8	13.0±1.8	13.5±1.7	14.7±1.5	CA†: e CB†: b,c,d
Wrist Flexors	13.6±2.3	13.5±2.1	13.4±2.1	14.5±1.5	12.9±2.4	13.7±2.2	14.5±2.6	14.7±1.6	CA: - CB†: c
Wrist Extensors	21.1±1.2	18.5±0.9	20.6±1.1	18.5±2.3	18.8±4.7	18.9±4.7	22.1±3.2	21.8±3.0	CA: - CB†: b,e

$^{\alpha}$indicates significant interaction effect (group x fencing distance), $p<0.05$

*indicates significant difference (main effect: group) between Category A and Category B fencers, $p<0.05$

†indicates significant difference (main effect: fencing distance), a: Distance_100 vs Distance_105; b: Distance_100 vs Distance_110; c: Distance_100 vs Distance_115; d: Distance_105 vs Distance_110; e: Distance_105 vs Distance_115; f: Distance_110 vs Distance_115

Table 6.19 Difference in Integrated EMG values (% MVIC) between the Category A and the Category B fencers at Distance_105, Distance_110 and Distance_115

Muscle	Difference in Integrated EMG between the Category A and the Category B fencers at different distance								
	Distance_105	Confidence Interval	p-value	Distance_110	Confidence Interval	p-value	Distance_115	Confidence Interval	p-value
Upper Trapezius	-	-	-	9.2	5.5 to 13.0	<0.001	15.0	10.9 to 19.2	<0.001
Infraspinatus	2.2	0.1 to 4.2	0.041	8.8	5.7 to 12.0	<0.001	12.0	9.1 to 14.9	<0.001
Anterior Deltoid	-	-	-	10.1	6.2 to 14.0	<0.001	20.9	16.3 to 25.4	<0.001
Mid Deltoid	3.6	0.25 to 6.9	0.036	11.4	9.2 to 13.6	<0.001	20.8	17.6 to 24.0	<0.001

6.3.2.4 Cross correlation of fencing arm muscles

Moderate to good cross correlations (R) were found in the motor patterns for Category A and B (Table 6.20). No significant interaction effect of group by fencing distance to R values was found, $p>0.05$. A highly synchronized upper limb muscular work was identified during the lunge attack.

The time lag (as expressed in percentage lunge cycle) amongst the different paired-muscle groups showed significant interaction group by distance effects only in the combination of BIC with the proximal shoulder muscles (i.e. UT, INF, MID and ANT). There were significant interaction effects for BIC-UT ($F_{3,84}=3.613$, $p=0.017$), BIC-INF ($F_{3,84}=14.099$, $p<0.001$), BIC-MID ($F_{3,84}=16.630$, $p<0.001$) and BIC-ANT ($F_{3,84}=14.923$, $p<0.001$). Time lag for BIC-UT, BIC-INF, BIC-MID and BIC-ANT in Category B progressively increased at longer fencing distance for within group effect; suggesting that the muscle patterns of the BIC muscles in Category B fencers tends to activate earlier than the proximal shoulder muscles as distance increased (Table 6.20). Such phenomenon was not observed in Category A fencers; no significant within-group effect was detected for the time lag of BIC-UT, BIC-INF, BIC-MID and BIC-ANT at Category A fencers. Post-hoc analysis revealed significant differences ($p<0.05$) in the time lag between Category A and Category B fencers at all fencing distances in BIC-UT, BIC-INF, BIC-MID and BIC-ANT

(Table 6.20).

6.3.2.5 Overall EMG results

All EMG results were summarized in Table 6.21 and Table 6.22. Table 6.21 illustrated the differences between Category A and B fencers at different fencing distances whereas Table 6.22 showed the within-group differences for Category A and B fencers at various fencing distances.

Table 6.20 Mean inter-muscle cross-correlation coefficients (R) and time lag (% lunge cycle) of the Category A and B fencers at different fencing distances

	Distance_100		Distance_105		Distance_110		Distance_115		Within-group difference	
	R	Lag (%)	R	Lag (%)	R	Lag (%)	R	Lag (%)	R	Lag (%)
UT and INF										
Category A	0.89±0.04	8.27±6.68	0.88±0.03	3.13±5.83	0.88±0.02	0.33±1.18	0.86±0.03	1.67±4.58	-	-
Category B	0.93±0.04	1.00±2.02	0.84±0.11	-1.60±6.93	0.84±0.12	-5.53±9.55	0.86±0.08	-3.27±4.50	-	-
UT and ANT										
Category A	0.93±0.04	0.04±3.91	0.92±0.03	0.13±1.96	0.92±0.03	-0.07±1.44	0.89±0.05	1.73±2.34	-	-
Category B	0.92±0.04	3.93±3.95	0.85±0.09	3.73±9.90	0.87±0.07	3.73±4.38	0.86±0.08	0.33±4.24	-	-
UT and MID										
Category A	0.89±0.04	13.93±9.18	0.87±0.04	7.20±9.57	0.85±0.05	5.07±9.98	0.84±0.06	3.73±6.51	-	-
Category B	0.93±0.03	4.87±4.87	0.86±0.09	-1.27±8.73	0.86±0.09	-2.20±5.20	0.86±0.06	3.93±6.92	-	-
UT and BIC $^{\alpha}$										
Category A	0.90±0.04	**9.20±5.47**	0.88±0.07	**4.40±3.18**	0.83±0.05	**3.60±3.16**	0.87±0.06	**-0.53±11.61**	-	†c
Category B	0.88±0.03	**-0.60±0.01***	0.84±0.08	**-10.93±1.61***	0.85±0.07	**-17.93±2.40***	0.87±0.05	**-18.4±3.41***	-	†a,b,c,d
UT and TRI										
Category A	0.81±0.04	15.80±11.10	0.80±0.05	20.40±2.90	0.79±0.04	13.87±13.29	0.80±0.05	13.49±12.79	-	-
Category B	0.75±0.11	12.67±12.69	0.75±0.08*	9.67±11.70	0.81±0.09	6.07±9.64	0.80±0.06	13.60±8.06	-	-
UT and WF										
Category A	0.79±0.09	19.47±13.85	0.78±0.07	19.70±12.10	0.73±0.05	23.80±10.47	0.73±0.05	13.40±15.34	-	-
Category B	0.75±0.11	4.93±18.17	0.75±0.11	3.93±15.66	0.79±0.11	4.73±10.75	0.79±0.08*	11.73±6.54	-	-
UT and WE										
Category A	0.79±0.05	12.60±11.44	0.78±0.04	6.00±10.33	0.79±0.03	-0.73±3.63	0.79±0.05	2.93±11.07	-	-
Category B	0.78±0.09	6.20±7.29	0.76±0.06	1.07±7.79	0.73±0.06	0.07±2.02	0.70±0.06	-1.27±18.00	-	-
INF and ANT										
Category A	0.95±0.02	-4.87±4.63	0.94±0.02	-3.33±3.52	0.94±0.02	-1.53±2.50	0.92±0.05	-1.87±2.72	-	-
Category B	0.94±0.03	1.00±2.83	0.95±0.02	0.07±1.28	0.93±0.04	0.33±2.80	0.95±0.02*	1.00±1.41	-	-
INF and MID										
Category A	0.95±0.03	1.87±2.47	0.96±0.02	0.47±1.73	0.96±0.01	-0.93±1.91	0.96±0.02	-0.20±1.08	-	-
Category B	0.93±0.04	2.07±2.87	0.95±0.02	1.40±2.26	0.94±0.04	0.47±0.83	0.95±0.02	1.00±1.56	-	-

INF and BIC$^{\alpha}$										
Category A	0.88±0.04	**2.20±6.98**	0.87±0.06	**1.40±7.24**	0.87±0.05	**2.20±5.82**	0.84±0.05	**3.87±3.70**	-	-
Category B	0.88±0.04	**-2.53±0.93***	0.89±0.03	**-6.20±1.20***	0.86±0.05	**-10.60±2.06***	0.89±0.02	**-17.07±4.56***	-	†b,c,e,f
INF and TRI										
Category A	0.91±0.03	1.73±3.71	0.89±0.03	2.87 + 5.19	0.92±0.02	0.07±1.71	0.90±0.03	1.53±3.23	-	-
Category B	0.78±0.09	7.60±6.42	0.74±0.10	9.07±12.14	0.72±0.06	6.33±10.27	0.78±0.07	4.13±5.06	-	-
INF and WF										
Category A	0.76±0.09	5.07±5.80	0.79±0.06	3.47±4.91	0.78±0.05	2.07±4.25	0.78±0.05	5.53±8.39	-	-
Category B	0.74±0.13	4.13±10.78	0.70±0.08	5.80±15.27	0.69±0.04	4.93±12.42	0.71±0.06	8.27±8.69	-	-
INF and WE										
Category A	0.87±0.04	0.53±2.07	0.85±0.03	-0.20±0.78	0.85±0.03	0.07±1.34	0.82±0.04	2.40±3.02	-	-
Category B	0.85±0.06	1.20±4.54	0.80±0.07	1.67±3.52	0.78±0.05	0.13±1.96	0.77±0.05	0.00±4.29	-	-
ANT and MID										
Category A	0.92±0.04	2.33±4.73	0.93±0.03	2.80±5.05	0.93±0.03	-0.33±10.48	0.91±0.06	3.33±5.95	-	-
Category B	0.95±0.02	1.07±2.02	0.97±0.01	0.53±0.92	0.96±0.01	0.20±0.86	0.96±0.01	0.00±0.38	-	-
ANT and BIC$^{\alpha}$										
Category A	0.88±0.04	**5.93±4.62**	0.90±0.05	**5.93±3.92**	0.88±0.05	**4.07±5.36**	0.87±0.07	**5.13±3.70**	-	-
Category B	0.91±0.05	**-4.33±1.02***	0.91±0.04	**-9.20±1.44***	0.88±0.03	**-15.73±3.39***	0.90±0.03	**-20.27±3.17***	-	†a,b,c,d,e
ANT and TRI										
Category A	0.76±0.04	9.80±5.05	0.78±0.02	11.87±5.51	0.76±0.03	9.87±7.27	0.74±0.05	5.67±6.88	-	-
Category B	0.78±0.09	4.47±6.01	0.74±0.08	9.53±6.58	0.76±0.07	9.07±6.20	0.79±0.06	6.53±6.23	-	-
ANT and WF										
Category A	0.74±0.09	9.67±10.48	0.76±0.06	10.13±9.20	0.75±0.06	12.07±8.52	0.74±0.05	9.00±11.43	-	-
Category B	0.74±0.10	0.40±8.90	0.68±0.09	6.13±7.95	0.72±0.04	6.53±7.32	0.73±0.05	6.13±6.16	-	-
ANT and WE										
Category A	0.80±0.04	3.33±5.95	0.82±0.03	3.27±5.66	0.81±0.04	2.40±4.76	0.78±0.04	5.00±8.09	-	-
Category B	0.83±0.06	2.93±3.37	0.80±0.07	1.93±5.60	0.79±0.06	4.27±5.75	0.76±0.04	0.87±4.26	-	-
MID and BIC$^{\alpha}$										
Category A	0.88±0.05	**0.47±4.55**	0.87±0.07	**-0.13±6.22**	0.90±0.05	**1.87±6.33**	0.87±0.05	**1.93±9.25**	-	-
Category B	0.90±0.04	**-7.00±1.69***	0.91±0.03	**-8.33±1.22***	0.88±0.04	**-16.40±2.12***	0.88±0.04	**-23.33±3.75***	-	†b,c,d,e
MID and TRI										
Category A	0.81±0.04	0.47±1.92	0.81±0.03	1.07±2.66	0.82±0.03	1.93±2.94	0.82±0.02	1.73±3.28	-	-
Category B	0.79±0.10	2.00±2.75	0.75±0.09	2.60±6.72	0.75±0.08	1.53±4.10	0.81±0.08	1.13±3.74	-	-

MID and WF										
Category A	0.78±0.07	4.13±5.40	0.79±0.06	4.60±8.89	0.81±0.06	6.73±5.84	0.79±0.05	5.53±8.39	-	-
Category B	0.73±0.10	2.00±7.90	0.69±0.08	6.87±13.07	0.72±0.05	1.47±6.93	0.73±0.05	2.33±5.95	-	-
MID and WE										
Category A	0.80±0.04	-0.47±2.72	0.83±0.03	0.87±2.42	0.83±0.04	3.33±4.29	0.83±0.04	2.53±3.44	-	-
Category B	0.84±0.07	1.40±4.93	0.80±0.06	0.93±3.54	0.81±0.05	-0.07 ±1.10	0.77±0.05	-0.60±2.35	-	-
BIC and TRI										
Category A	0.88±0.04	9.40±7.32	0.90±0.06	14.93±4.48	0.89±0.05	14.27±4.88	0.86±0.04	12.60±4.76	-	-
Category B	0.79±0.09	11.93±9.52	0.77±0.03	20.60±9.90	0.78±0.05	28.93±7.85	0.79±0.04	28.93±3.69	-	-
BIC and WF										
Category A	0.81±0.08	14.27±10.00	0.87±0.06	20.60±5.76	0.84±0.06	19.67±4.72	0.82±0.06	17.93±7.30	-	-
Category B	0.72±0.09	7.87±18.17	0.73±0.06	19.67±11.48	0.76±0.03	26.80±12.48	0.76±0.06	31.47±5.13	-	-
BIC and WE										
Category A	0.80±0.04	11.07±5.09	0.77±0.06	9.27±6.03	0.78±0.07	15.47±8.83	0.81±0.07	16.80±5.94	-	-
Category B	0.82±0.07	8.33±7.14	0.75±0.05	14.93±7.78*	0.76±0.04	13.4±9.93	0.72±0.04	10.00±15.80	-	-
TRI and WF										
Category A	0.84±0.08	4.60±4.61	0.86±0.06	2.53±4.94	0.86±0.05	4.47±6.29	0.86±0.06	1.80±5.28	-	-
Category B	0.85±0.10	-2.47±9.01	0.90±0.06	-1.47±2.17	0.90±0.03	-1.67±1.99	0.87±0.06	-3.13±2.97	-	-
TRI and WE										
Category A	0.90±0.04	0.40±2.16	0.86±0.04	-0.07±1.87	0.87±0.03	2.60±4.88	0.89±0.03	4.00±4.28	-	-
Category B	0.87±0.08	-1.33±2.55	0.83±0.11	-1.33±2.61	0.82±0.08	-0.67±3.06	0.81±0.05	-1.47±2.53	-	-
WF and WE										
Category A	0.81±0.07	-0.60±5.97	0.78±0.08	-3.60±4.98	0.81±0.06	-0.27±3.173	0.82±0.07	-2.93±6.96	-	-
Category B	0.81±0.10	-0.13±8.40	0.80±0.10	-1.07±4.73	0.84±0.07	-0.87±4.61	0.76±0.06	-3.73±4.04	-	-

+ve: movements in-phase; -ve: movements out-of-phase

$^{\alpha}$indicates significant interaction for time lag (group x fencing distance), $p<0.05$

$^{\#}$indicates significant interaction for cross-correlation (group x fencing distance), $p<0.05$

*indicates significant difference (main effect: group) between Category A and Category B fencers, $p<0.05$

†indicates significant difference (main effect: fencing distance), a: Distance_100 vs Distance_105; b: Distance_100 vs Distance_110; c: Distance_100 vs Distance_115; d: Distance_105 vs Distance_110; e: Distance_105 vs Distance_115; f: Distance_110 vs Distance_115

Table 6.21 Summary of the between-group differences of the EMG variables in Category A and Category B fencers at various fencing distances

EMG variable (unit)	Category A vs Category B	Upper Trapezius	Infraspinatus	Ant-deltoid	Mid-deltoid	Biceps	Triceps	Wrist flexors	Wrist extensors
EMG onset (% lunge cycle)	Distance_100	-	-	-	-	-	-	-	-
	Distance_105	-	-	-	**CB ↑***	-	-	-	-
	Distance_110	-	**CB ↓***	**CB ↓***	**CB ↓***	**CB ↓***	**CB ↓***	**CB ↓***	**CB ↓***
	Distance_115	-	**CB ↓***	-	**CB ↓***	**CB ↓***	**CB ↓***	**CB ↓***	**CB ↓***
Occurrence of peak EMG (% lunge cycle)	Distance_100	-	-	-	-	-	**CB ↑***	-	-
	Distance_105	-	-	-	-	**CB ↓***	**CB ↓***	**CB ↓***	**CB ↓***
	Distance_110	-	-	-	-	-	**CB ↓***	**CB ↓***	**CB ↓***
	Distance_115	-	-	-	-	-	**CB ↓***	**CB ↓***	**CB ↓***
Peak EMG (%MVIC)	Distance_100	-	-	-	-	-	-	-	-
	Distance_105	-	**CB ↓***	**CB ↑***	-	-	-	-	-
	Distance_110	**CB ↑***	**CB ↑***	**CB ↑***	**CB ↑***	-	-	-	-
	Distance_115	**CB ↑***	**CB ↑***	**CB ↑***	**CB ↑***	-	-	-	-
Integrated EMG (%MVIC)	Distance_100	-	-	-	-	-	-	-	-
	Distance_105	-	**CB ↑***	-	**CB ↑***	-	-	-	-
	Distance_110	**CB ↑***	**CB ↑***	**CB ↑***	**CB ↑***	-	-	-	-
	Distance_115	**CB ↑***	**CB ↑***	**CB ↑***	**CB ↑***	**CB ↑***	-	-	-

*p<0.05

Table 6.22 Summary of the within-group differences of the EMG variables in Category A and Category B fencers at various fencing distances

EMG variable	Comparison	UT		INF		ANT		MID		BIC		TRI		WF		WE	
		CA	CB	CA	CB	CA	CB	CA	CB	CA	CB	CA	CB	CA	CB	CA	CB
EMG onset (% Lunge cycle)	Distance_100 Vs 105	-	-	*	*	*	*	-	-	*	-	-	-	-	-	-	*
	Distance_100 Vs 110	*	*	*	*	*	*	*	*	*	*	*	*	-	*	-	*
	Distance_100 Vs 115	-	*	*	*	*	*	-	*	*	*	-	*	-	*	-	*
	Distance_105 Vs 110	-	*	-	*	-	*	-	*	-	*	-	*	-	-	-	*
	Distance_105 Vs 115	-	*	-	*	-	*	-	*	-	*	-	*	-	-	-	*
	Distance_110 Vs 115	-	-	-	*	-	-	-	-	-	-	-	-	-	-	-	-
Occurrence of peak EMG (% lunge cycle)	Distance_100 Vs 105	-	*	-	*	-	-	-	-	-	*	*	*	*	*	*	*
	Distance_100 Vs 110	-	-	-	*	-	-	-	-	-	*	-	*	*	*	*	*
	Distance_100 Vs 115	-	-	-	-	-	-	-	-	-	*	-	*	*	*	*	*
	Distance_105 Vs 110	-	-	-	-	-	-	-	-	-	*	*	*	-	*	-	*
	Distance_105 Vs 115	-	-	-	-	-	*	-	-	-	-	-	*	-	*	-	*
	Distance_110 Vs 115	-	-	-	*	-	-	-	-	*	-	-	-	-	-	-	-
Peak EMG (%MVIC)	Distance_100 Vs 105	*	*	*	*	-	-	-	*	-	-	-	-	-	-	*	*
	Distance_100 Vs 110	*	*	*	*	-	*	-	*	-	-	*	-	-	-	*	-
	Distance_100 Vs 115	-	*	*	*	-	*	-	*	-	-	-	-	-	-	-	*
	Distance_105 Vs 110	-	*	*	*	-	*	-	*	-	-	*	*	-	-	-	-
	Distance_105 Vs 115	-	*	*	*	-	*	-	*	-	-	-	-	-	*	-	*
	Distance_110 Vs 115	-	*	*	-	-	-	-	-	*	-	-	-	-	*	-	-
Integrated EMG (%MVIC)	Distance_100 Vs 105	-	-	-	*	-	*	-	*	-	*	-	-	-	-	-	-
	Distance_100 Vs 110	-	*	-	*	-	*	-	*	-	*	-	*	-	-	-	*
	Distance_100 Vs 115	-	*	-	*	-	*	-	*	*	*	-	*	-	*	-	-
	Distance_105 Vs 110	-	*	-	*	-	*	-	*	-	-	-	*	-	-	-	-
	Distance_105 Vs 115	-	*	-	*	-	*	-	*	-	-	*	-	-	-	-	*
	Distance_110 Vs 115	-	*	-	*	-	*	-	*	-	-	-	-	-	-	-	-

***p<0.05**

6.4 Discussion

This study, to our knowledge, was the first to describe the upper limb kinematics and kinetic features in wheelchair fencing. As one of the fundamental techniques to score, lunge attack in wheelchair fencing was examined. A unique upper limb movement and motor recruitment pattern during lunge attack was recorded. Category A and B wheelchair fencers, despite their difference in physical disabilities, showed a common recruitment pattern during the lunge attack at short fencing distance of Distance_100 and Distance_105. There were, however, significant differences in both kinematic and EMG outcomes between Category A and Category B fencers; included significantly longer lunge duration, larger angular displacement, lower peak linear and angular velocity, reduced shoulder-elbow joint coordination, earlier recruitment of muscles (especially with biceps) and a higher activation level of the shoulder muscles in Category B fencers at Distance_110 and Distance_115. The lack of active lower limb and trunk control, Category B fencers adopt a different kinematic and motor recruitment pattern as they performed the lunge at longer fencing distances.

6.4.1 Lunge attack description in wheelchair fencing

6.4.1.1 Kinematics of the lunge attack motion

The execution of a lunge attack in wheelchair fencing involves a powerful and coordinated shoulder abduction, flexion, internal rotation, elbow extension, forearm pronation, wrist flexion and ulnar-deviation. A relatively large shoulder and elbow angular movement in the coronal plane, i.e. a lateral reaching movement, was quantified in the present study. As the attack distance increased, a corresponding increase in angular displacement of the upper limb joints was found. Specifically, the shoulder flexion, abduction and rotation angular displacement reach up to 107.2°, 79.7° and 69.0° respectively in Category A, and 124.3°, 91.9° and 90° respectively in Category B.

The lunge attack for both Category A and B fencers was also characterized by moderate to high cross-correlation coefficients between shoulder flexion and elbow extension (R ranges from 0.87-0.91), shoulder abduction and elbow extension (R ranges 0.74-0.86) and elbow extension and forearm pronation (R ranges from 0.94-0.98), suggesting the close coupling and the importance of these synergic movements for performing the lunge attack.

Compared to other available kinematic data in literature, the able-bodied fencers had substantially higher peak weapon velocity for lunge attack than that of wheelchair fencers. The maximum linear horizontal velocity of wheelchair fencing measured in this study was 3.30 m/s in Category A and 2.56 m/s in Category B fencers. Hasson, et al. (1998) and Lopez, et al. (2007) recorded up to 3.91 to 4.02 m/s in able-bodied fencers; approximately 1.5 times faster than that of the wheelchair fencers. In able-bodied fencing, high weapon velocity is attained mainly through the lower limbs. The highly-coordinated actions of plantar-flexors, knee extensors and hip abductors over the fencers' trail leg generate a powerful horizontal thrust in the attack direction (Gebhard, 1981; Morris, et al., 2011; Gholipour, 2008; Suchanowski, 2011). Without the power generated by the lower limbs, wheelchair fencers rely solely on their trunk and fencing arm to perform the lunge attacks, resulting in a less powerful execution.

Despite a lower performance in linear speed generation in wheelchair fencers, the angular velocity for some of the upper limb joint motion in wheelchair fencing was higher than their able-bodied counterpart. In Frere, et al. studies (2011), the elbow extension peak angular velocity during lunge attack in 8 expert able-bodied fencers ranged from 305.9-656.8°/s; which was comparable to our finding of 598.2°/s. However, the shoulder flexion peak angular velocity in our study was much higher than that of documented able-bodied fencers: 430.3-655.1°/s versus 1065.7°/s. The exceptionally high angular velocity may induce a higher risk of injury to wheelchair fencers to sustain from various shoulder disorders; that substantiate the predominance of shoulder injuries in wheelchair fencers as found in our previous epidemiological study.

6.4.1.2 EMG activity during lunge attack motion

6.4.1.2.1 Motor recruitment sequence

Wheelchair fencers exhibited a v ery consistent proximal-to-distal motor recruitment sequence of their upper limb during lunge attack at all fencing distances. This sequence also applied to the occurrence of peak EMG. Similar motor recruitment sequence was reported in other ballistic action sports such as the karate punch and Kung Fu strike (Neto & Magini, 2008; VencesBrito, et al., 2011) where foot placement was fixed to the ground during the actions. In these studies, muscle recruitment sequence initiates from anterior deltoid, follows by biceps, triceps and then the distal muscles of pronator teres and brachioradialis. Such proximal-to-distal sequence, according to kinetic chain theory, is important in effective speed generation and the distal segments acceleration to achieve higher force and power in sport activities (Hirashima, et al. 2002). Unfortunately, comprehensive assessment about the motor recruitment pattern in able-bodied fencing is lacking. Direct comparison between the motor recruitment pattern between the able-bodied and wheelchair fencers is not feasible at this stage.

6.4.1.2.2 Motor activation level

A higher motor activation level (i.e. iEMG and peak EMG) was recorded in the proximal shoulder muscles (UT, INF, ANT, MID and BIC: peak EMG ranges from 15-25%MVIC and iEMG ranges from 21-47%MVIC at Distance_100) as compared to the upper arm and forearm muscles (BIC, TRI, WF and WE: peak EMG ranged from 5-10% MVIC and iEMG ranges from 13-10%MVIC at Distance_100) in both Category A and Category B fencers; indicating the major roles of the shoulder muscles for lunge attack.

The EMG muscle activity of lower leg in able-bodied fencers was much higher than that of the fencing arm (Suchanowski, et al., 2011; Williams, et al., 2000). However, the EMG signals in these studies were not amplitude normalized and a direct comparison with the present findings is not appropiate.

6.4.1.2.3 Roles of proximal and distal upper limb muscles

Shoulder:

Anterior-Deltoid, Mid-deltoid, upper trapezius and infraspinatus always exhibit the highest (Peak EMG amplitude from 15-27 %MVIC in Category A and 14-40% MVIC in Category B) and earlier SEMG (EMG onset from 34%-41% lunge cycle in Category A and 30%-40% lunge cycle in Category B) activity (Table 6.10 and 6.14), illustrating their major roles as the activators to initiate the quick lunge execution. During the early phase of lunge attack, the strong activation of deltoid elevates the humerus and brings up the briskly shoulder abduction and flexion. Upper trapezius, the key regulator to the upward rotation of scapula, activates simultaneously to coupling the explosive shoulder abduction. Infraspinatus acts as the active stabilizer to hold the humeral head within the glenohumeral joint and facilitates the deltoid muscle to maintain the flexion-abduction positions (Kelly, et al., 2002; Kronberg, et al., 1990; Magarey & Jones, 2003). Frere, et al. (2011) also suggested the role of infraspinatus to offset the upward pull generated by the anterior deltoid muscle. The high cross-correlation and the negligible time lag amongst the four proximal shoulder muscles supported the synchronization of these muscles for executing the lunge attack motion. The close coupling of the upper trapezius and infraspinatus regulates the scapula and humeral head respectively so that a powerful shoulder elevation activated by anterior- and mid-deltoid was executed during the lunge attack motion.

Elbow:

Biceps (EMG onset from 34%-44% lunge cycle in Category A and 30%-42% lunge cycle in Category B) is always activated before triceps (EMG onset from 55%-61% lunge cycle in Category A and 48%-65% lunge cycle in Category B) in all wheelchair fencers (Table 6.10). However, when matching our kinematic analysis to the muscle activity, there was no elbow movement being identified, indicating that the biceps has two roles during the lunge attack. First, when upper limb started to elevate, the weight of the weapon and arm caused the biceps to contract isometrically to stabilize the elbow joint even there was no elbow movement. Second, when the fencer extended the elbow to hit the target, the triceps contracted rapidly and the biceps work antagonistically and eccentrically, attempting to control the elbow movement. Our findings are coherent with VencesBrito, et al. (2011) who shown that the earlier recruitment of biceps to hold the arm before the activation of triceps for karate punch.

Wrist:

The wrist flexors and extensors usually recruit at a similar onset time (EMG onset from 71%-75% lunge cycle in Category A and 60%-77% lunge cycle in Category B) and trigger a w eak EMG activity (Peak EMG amplitude from 7-12 %MVIC in Category A and 5-14% MVIC in Category B) prior to impact during the lunge attack; suggested the muscles do not contribute in any speed development but to align and stabilize the wrist and hand before collision (Table 6.10). These findings concur to other motion studies suggesting that shoulder and elbow are for speed generation in forward pointing task whereas the wrist and hand are mainly responsible for orientation and precision of pointing (Kaminski, et al., 1995). The co-contraction of wrist flexors and extensors may assist in the dynamic wrist stability and protect the upper limb structures from injury by the high impacting force.

6.4.2 Movement and motor adaptation in Category B fencers during lunge attack at longer fencing distance

While the kinematic features and EMG characteristics are remarkably similar between the Category A and Category B fencers at short fencing distances, significant differences were found at longer fencing distances. The following section will discuss the kinematics, EMG, inter-joint coordination and inter-muscle activation differences between the two fencing groups.

6.4.2.1 Kinematic differences between Category A and Category B fencers

The Category B fencers differed from the Category A fencers in three specific areas in related to kinematics when executing the lunging action with increase fencing distance: i) slower linear horizontal and angular velocity at shoulder and elbow joint; ii) larger shoulder angular displacement; and iii) a deprivation of the shoulder-elbow joint coordination at longer fencing length.

The compromised trunk control in Category B fencers created different kinematics. Toyoshima, et al. (1974) found that there was 36.5% drop in peak ball velocity during normal overhead throw when the trunk and lower limb movement of the subjects were restricted. Supportive finding of Alexander (1991) also demonstrated that restriction of trunk motion could significantly reduce the maximal ball throw velocity up to 50%. Disabled athletes including wheelchair tennis (Reid, et al., 2007), wheelchair basketball (Malone, 2002; Nunome, 2002) and seated discus throw (Chow, et al., 1999; 2003) with poor trunk control also demonstrates a lower upper limb performance and kinematic outcomes as compared to the able-bodied counterparts.

Apart from being not able to reach a higher peak velocity, wheelchair fencers having poor trunk control also found to perform with some adapted patterns as compensation. A larger shoulder angular displacement was noted in Category B fencers when they execute the lunge at longer distance of Distance_110 and Distance_115. A 14.4% and 16.0% larger shoulder flexion angular displacement was found in Category B fencers at Distance_110 (114.2±8.8°) and Distance_115 (124.3±8.1°) (Table 6.5). For shoulder abduction, a similar increase of 13.2% and 15.3% of the shoulder abduction range at Distance_110 (84.8±7.3°) and Distance_115 (91.9±5.2°) (Table 6.5). Similar observations were reported by previous studies on wheelchair athletes and spinal cord injured subjects. Biomechanical studies on manual wheelchair propulsion reported by Boninger, et al. (2005), Koontz, et al. (2002) and Newsam, et al. (1999) illustrated higher-level spinal-cord-injured (SCI) subjects produce a larger range of motion of the upper limb joints than those with lower-level SCI. The compromised trunk movement expectedly relies on m ore upper limb movement adaptations. The reduction in peak velocity (both linear horizontal and angular velocity) and the elevation in shoulder joint angular displacement in Category B fencers were more noticeable at longer distances.

Category B fencers had also adapted a d ifferent inter-joint coordination over their shoulder and elbow joint as they lunge at longer distances. The cross-correlation analysis showed strong correlation between shoulder and elbow joint in all fencing groups, suggesting the good dynamic coupling strategy between the two joints during lunge attack. As fencing distances increased, Category B fencers exhibited a significant drop in the cross-correlation coefficients of shoulder and elbow and a larger lag. This altered joint coordination is possibly due to the adjusted motor recruitment pattern of torso, scapular and shoulder muscles during the reaching movement at the longer fencing distance. Electromyographic analyses performed by Potten, et al. (1997) and Seelen, et al. (2001) pointed out that spinal cord injury subjects made alternative use of the non-postural muscles including dorsal scapular muscles, latissimus dorsi and trapezius muscle for balance during forward reaching tasks. Such adaptation is more obvious in subjects with higher level spinal cord lesions (Potten, et al., 1999; Seelen, et al., 2001). The increased demand of the residual innervated muscles over the dorsal scapular region may affect the motor activity of the muscles around the scapulohumeral region, leading to the altered shoulder coordination that was found in this study.

Levin, et al. (2002) reported that the reaching distance has significant impact on the spatiotemporal coordination of body segments during reaching movement. Reaching within arm's length only requires elbow and shoulder movements; while targets beyond arm's length require also trunk movements. In shorter lunge distance, there was no significant difference of maximum lunge velocity and maximum lunge angle between the Category A and Category B fencers, which is consistent with the trunk kinematic study by Fung, et al. (2013). In Fung's study, the trunk velocity and range of motion of the 9 Category A and 5 Category B fencers showed no significant difference. However, the testing distance in Fung's study was set only at the normalized fencing distance. In contrast, our study analyzed upper limb kinematic at a longer fencing distance; reflecting a genuine condition that is more compatible to valid competition or practice simulation.

6.4.2.2 Electromyographic differences between Category A and Category B fencers

6.4.2.2.1 Motor activation level

Kelly, et al. (2002) classified the percentage values of MVIC into three categories: minimal activation, below 35% MVIC; moderate activation, 35% to 70% MVIC; and maximal activation, more than 70% MVIC. The overall peak EMG level for wheelchair fencing was generally low as compared to other upper limb sport events such as the football throw and volleyball spike (Table 6.23).

Table 6.23 Muscle activation level (%MVIC) for various upper limb sports

Peak EMG value (% of MVIC)	EMG Technique	Supraspinatus	Infraspinatus	Anterior Deltoid	Mid Deltoid
Wheelchair fencing	Surface EMG				
Lunge attack					
(Distance_100)					
Category A fencer		-	14±3	22±5	19±3
Category B fencer		-	14±1	25±11	22±3
(Distance_115)					
Category A fencer		-	29±6	27±6	20±5
Category B fencer		-	34±6	40±3	29±3
Volleyball[ε]	Needle EMG				
Serve		45±13	39±21	42±17	--
Spike		71±31	60±17	58±26	--
Football throw[‡]	Surface EMG	87±43	86±33	49±14	48±19

‡ Kelly, et al., 2002
ε Rokito, et al., 1998

However, as far as the shoulder joint alone is concerned, it is notably that the peak EMG and iEMG escalated as fencing distance increases, particularly in Category

B fencers. Peak EMG for upper trapezus (Category A: 20.5±5.3%MVIC; Category B:30.3±8.6%MVIC), infraspinatus (Category A:28.6±6.1%MVIC; Category B: 34.2± 6.0%MVIC), anterior-deltoid (Category A: 26.1±3.3%MVIC; Category B: 39.5± 2.9%MVIC) and mid-deltoid (Category A: 18.0±2.5%MVIC; Category B: 29.4± 2.8%MVIC) at Distance_115 was found to increase by 38%, 11%, 19% and10% respectively in Category A fencers, and 109%, 153%, 55% and 35% respectively in Category B fencers as compared to that at Distance_100. The total muscle work for lunge attack at Distance_115 was slightly higher in Category A fencers and substantially higher in Category B fencers. The iEMG amplitudes for upper trapezius (Category A: 39.0±4.2%MVIC; Category B: 54.1±4.8%MVIC), infraspinatus (Category A: 25.8%MVIC; Category B: 37.8±4.8%MVIC), anterior-deltoid (Category A: 46.7±2.9%MVIC; Category B: 67.6±8.6%MVIC) and mid-deltoid (Category A: 25.8%MVIC; Category B: 46.5±4.8%MVIC) showed to have an average of 0-7.4% and 56.5-72.8% increase in iEMG in Category A and Category B fencers respectively. These findings concur with the results from previous studies that individuals with poor trunk control exhibit higher upper limb muscle activities than those with good trunk control (Do, et al., 1985; Mulroy, et al., 2004; Potten, et al., 1999; Seelen, et al., 1997). Seelen & Vuurman (1991) and Seelen, et al. (1997) used the bimanual forward-reaching movement to study postural control in SCI people with lesion between T2 and T12. They found that paraplegic subjects used latissimus dorsi and trapezius muscle to control their sitting balance. In another study, Dean, et al. (1999) showed that strong muscle activity of the lower extremities were observed when subjects performed the functional forward reach up to 140% of their arm length in able-bodied and incomplete SCI subjects. As the lunge distance increases, the torque between the trunk and the fencing arm increases proportionally as a result of longer moment arm of the weapon. Higher force around scapulothoracic and glenohumeral muscles is required to elevate the arm and maintain postural support; thus significantly produce higher peak EMG in both prime movers and stabilizers around shoulder. In the absence of trunk innervations and normal trunk synergistic stabilization (i.e. Category B), wheelchair fencers' upper extremity fencing motions were performed in isolation, thus requiring a higher muscular effort (i.e. 14-44% and 20-63% higher peak EMG amplitude for the proximal shoulder muscles at Distance_110 and Distance_115 respectively in Category B than the Category A fencers) as suggested by our results.

Although the muscle recruitment levels of the investigated upper limb muscles during lunge attack in wheelchair fencing were categorized as minimal, elite wheelchair fencers may undergo more than 40 hours of high-intensity training per week. This problem may be more prominent in Category B fencers as their fencing arm muscle activities are persistently higher than those of Category A fencers, especially at longer fencing distance (i.e. Distance_110 and Distance_115). The repetitive overloading of the shoulder muscles seen in current study plays a significant role in the etiology of such injuries among wheelchair fencers. The current findings not only improve the understanding of the potential causes of shoulder injuries in wheelchair fencers, but also provide empirical information for the design of more effective practice protocol for wheelchair fencers rehabilitating from existing injuries or wanting to minimize strain during routine practice.

6.4.2.2.2Earlier onset and earlier occurrence of peak EMG

The onsets of EMG for all upper limb muscles were kept steady across the different fencing distances in Category A fencers. However, in Category B fencers, an earlier onset of EMG was found as distance increased; indicating a possible adaptive strategy to maintain balance. We suggest, at short fencing distances (i.e. Distance_100 and Distance_105), sitting balance is generally not disturbed and both Category A and Category B fencers recruit their muscles in a usual way for lunge execution. As fencing distance increases, the lever arm inevitably increases due to the inclined trunk. Although it is speculative, Category B fencers may adapt an earlier recruitment of their upper limb muscles from proximal-to-distal sequence, to maintain balance.

Despite an earlier onset of EMG activity in Category B, it is interesting to note that Category B fencers was not prone to trigger an earlier peak EMG for accustoming the balance control at longer fencing distances. Rather, Category B fencers tend to substantially increase their peak EMG amplitude over proximal shoulder muscles so as to improve the shoulder stability. This strategy coincides with the findings of Seelen & Vuurman (1990) that an adaptive compensation of thoracoscapular muscles for maintaining the balance as the subjects were instructed to perform the forward reaching activity beyond the arm-length.

6.4.2.3 Inter-joint and inter-muscle coordination

This is the first study in wheelchair fencing to employ cross-correlation to study inter-joint and inter-muscle work, though the technique has previously been used to study the spine and hip in people with low back pain (Wong & Lee, 2004; Shum, et al., 2007). Cross-correlation analysis used in this study provides an objective method to evaluate the time-history kinematics and EMG patterns of the upper limb joints or muscles. In addition to the discrete data from the EMG time domain (i.e. onset and occurrence of peak) and amplitude domain (peak EMG), a comprehensive picture of the muscle activation could be obtained.

Regarding the coupling of joint movement, our results demonstrated a high inter-joint coordination existed between i) shoulder flexion and elbow extension; ii) shoulder abduction and elbow extension, and iii) elbow extension and forearm pronation for both Category A and Category B fencers. However, with increasing fencing distance, there was a gradual loss of inter-joint coordination between shoulder flexion and elbow extension and, shoulder abduction and elbow extension but not elbow extension and pronation in Category B fencers. The time lag between shoulder flexion and elbow extension correspondingly increased for both Category A and Category B fencers. At Distance_110 and Distance_115, the time lag was 5.34% and 4.17% higher in Category B than that of the Category A respectively. From the cross-correlation between the different fencing arm muscles, it was note that an earlier EMG profile of biceps muscle was observed when comparing to other proximal shoulder muscles in Category B fencers at all four fencing distances. It is postulated that the earlier biceps recruitment in Category B fencers may solicit earlier and work eccentrically to control the extension of elbow during lunge so as to reduce the lever arm during the lunge attack. The early biceps recruitment helps control the increase in moment arm and strategically minimizes the demand on postural balance in Category B fencers as they lunge at longer fencing distances.

Beside of the compatible onset and occurrence of peak EMG of the four key shoulder muscles (upper trapezius, anterior deltoid, MID-deltoid and infraspinatus) as discussed in the previous section, a high degree of similarity in the EMG profiles were observed amongst these muscles. High cross-correlation values ($R \geq 0.90$) with negligible lag (within 5% lunge cycle) were persistently found amongst the four muscles in both fencing groups at all fencing distances, suggesting the highly synchronized inter-muscle movements during the lunge attack.

As a summary, both Category A and B wheelchair fencers showed a common recruitment pattern during the lunge attack at short fencing distance. As distance increased, significantly longer lunge duration, larger angular displacement, lower peak linear and angular velocity, reduced shoulder-elbow joint coordination, earlier recruitment of muscles (especially with biceps) and a higher activation level of the shoulder muscles were found in both groups, more in Category B than Category A. Compared to the able-bodied counterparts, wheelchair fencers generated lower linear speed but higher angular velocity of their upper limb motions.

As fencing distance increased, Category B fencers had slower linear horizontal and angular velocity at shoulder and elbow joint, larger shoulder angular displacement, recruited muscles earlier and a deprivation of the shoulder-elbow joint coordination when compared to Category A fencers. Biceps showed an earlier recruitment comparing to other proximal shoulder muscles in Category B fencers regardless fencing distances due to its dual roles as stabilizer as well as controller for elbow extension. This change in recruitment pattern supports the kinetic chain principle: any change in recruitment pattern causing a break in the kinetic change, consequently lead to increased loading. The current results support the physical factors we proposed in our conceptual model included the force, posture, and velocity are all among the contributing factors to the performance of the fencing actions. Different from able-bodied fencers, the omission of footwork in wheelchair fencer limits the force generation from ground reaction. The kinetic chain of wheelchair fencers starts from the trunk and inevitably reduces the linear speed generation during the lunge attack. In order to compensate and acquire a higher speed, wheelchair fencers exert a larger angular speed over their shoulders. At shorter fencing distance, both Category A and B fencers performed similarly. As fencing distance progresses, the postural stability for both fencing groups was being challenged. Due to the loss of active trunk control, the kinetic chain in Category B is further disrupted. Category B fencers adopt a series of compensatory strategy including an earlier recruitment of upper limb muscles and a higher muscle work of the shoulder muscle to maintain the postural stability. Meanwhile, an earlier activation of the biceps help to control the increase in lever arm due to the thrustful arm elevation and elbow extension. To accomplish the lunge attack at longer fencing distance, Category B needs to exert a larger shoulder range but a significant lower linear and angular speed.

6.4.3 Clinical implications

The upper limb movement patterns identified in the present study may provide some insight to the understanding of the physical risk factors in wheelchair fencing and the prevalence of shoulder injuries in wheelchair fencers. Wheelchair fencers execute the powerful, briskly and repetitive attacks by a typical fencing pattern of shoulder flexion and abduction up to 100° to 120°. This physiological range, in combination with the shoulder internal rotation, may place stress to rotator cuff tendons and sub-acromial tissues (Hughes, et al., 2012; Martetschlager, et al., 2012). In addition to the higher torque exerted on the shoulder structures due to extended lever length of the extended arm and weapon, the shoulder structures could be more predisposed to cuff disorders and impingement problems.

For individuals with loss of active trunk control, the altered inter-joint coordination during lunge at longer distance may further increase the stress to their shoulders with truncated "kinetic chain". Findings of this study showed that wheelchair fencers (especially Category B) used altered kinematics over their shoulder complex. Such breakage in the kinetic chain by the loss of inter-joint coordination may put biomechanical disadvantages to the upper limb joints, implying a higher risk of upper limb injury. Wheelchair fencers, especially those with poor trunk control, are recommended to strength their residual innervated musculature and shoulder muscles, so as to optimize their shoulder kinematics for the protection of potential shoulder pathology.

Attention for prevention of injury should also be directed toward proper lunge mechanics coaching, proper rest during intensified trainings education and regulation of the number of lunges in wheelchair fencers. Prospective research on the effect of shoulder strength training for wheelchair fencing injury prevention is required.

In fact, our results indicate that Category B fencers have altered motor recruitment pattern and higher muscle activity. Coupling with the highly repetitive nature of the lunge action during wheelchair fencing, implying a higher muscular load is anticipated to exert onto the shoulder structure, thus increase the risk of developing shoulder pain/injury.

Understanding the muscle recruitment patterns of wheelchair fencers is important for enhancing athletes' performance, preventing their injury and improving the existing rehabilitation programs. Improving the strength of the key muscles for lunge i.e. deltoids, upper trapezius, infraspinatus and biceps in response to the higher muscular demand in wheelchair fencers with poor trunk control should be emphasized. Strength training at various fencing distances, particularly at long fencing length, may further improve the training specificity for the fencing arm muscles. As documented in our previous injury surveillance, shoulder muscle injuries, including biceps tendinitis, were predominance in Category B fencers. The implementation of eccentric training for biceps muscle in Category B fencers may be helpful to substantiate its role to stabilize the elbow during lunge attack and minimize the chance of injury. At the same time, trainers and therapists should also be cautious to preserve muscle balance around the glenohumeral region. As muscle activations during lunge attack proceed in a proximal-to-distal sequence, conditioning exercises should adhere to this progression for ensuring inter-muscular coordination.

Due to the adaptive role of the proximal glenohumeral muscles in stabilization, special attention to the training of core musculature linking the trunk and shoulder girdle should be implemented. Core stability exercises may augment the transfer of force from trunk to upper extremity. Neuromuscular training should be incorporated to optimize the motor recruitment and improve the strength of the residual innervated trunk muscles for the proximal trunk support.

Besides, a large angular displacement over the shoulder was required for lunge attack at a long distance. It is important that warm-up regime should include shoulder mobilization and flexibility exercise for improving the mobility of the fencers' shoulder joints prior to practice.

The concept of informing the athletes about the potential risk of a specific sport has been recently promoted in sports medicine (Webborn, 2012) and should be incorporated into the wheelchair fencing sport. Wheelchair fencers should understand the risk and make informed decision on t he participation of wheelchair fencing. Coaches, fencers and medical personnel should also be aware of the relative high motor demand and the vulnerability of shoulder among wheelchair fencers with poor trunk control. Any early sign of shoulder disorder in these athletes should be given appropriate medical assessment and intervention to prevent further injury and promote recovery.

6.4.4 Conclusion

Lunge action of the world-class wheelchair fencers was quantified and the kinematic and electromyographic features of wheelchair fencers with good- and poor- trunk control were compared. Wheelchair fencers with poor trunk control demonstrated a different kinematics with significantly lower peak horizontal and angular velocities, larger angular displacement and altered joint coordination over their shoulder and elbow joints as fencing distance increased. The altered kinematics might represent a unique shoulder movement strategy used by Category B fencers to compensate for their poor trunk control when they fenced at longer distances. This altered shoulder kinematics may induce excessive stress and result in higher incidence of shoulder injury in this group of fencers. However, due to the cross-sectional nature of the present study, the causal relation between such kinematic alternation and shoulder pain in Category B fencers was not established. Regarding the EMG activities, wheelchair fencers with poor trunk control were shown to have different motor pattern and a higher muscle activity over their shoulders. The increased muscle activity and the highly repetitive nature of lunge attack may predispose Category B fencers to high risk of shoulder disorders. Targeted strengthening program at longer fencing distances and stabilization exercises of proximal shoulder muscles are recommended to enhance the lunge performance and prevent injury. Further EMG studies that involve female wheelchair fencers, novice fencers, different level of disabilities and the investigation of the deeply-located scapulothoracic and scapulohumeral muscles may improve the understanding of the mechanism of injury or biomechanics of those subjects.

Results from the current exploratory study may provide useful information for better understanding of the possible upper limb injury in wheelchair fencers, especially for the Category B fencers who have poor trunk control. Additionally, the findings of this study may add new knowledge for establishing injury prevention program or rehabilitation strategies that are specific to wheelchair fencing.

Note: Part of this chapter was presented in the the 5th World Congress on Bioengineering, Aug 18-21, 2011; Tainan, Taiwan and the Student Conference on Sports Science, Rehabilitation and Medicine 2013, November 30, 2013

CHAPTER 7 – General Summary and Conclusion

7.1 Summary and findings

Wheelchair fencing has a long standing history in disabled sports. Due to the small number of participants, paucity of systematic injury research and the complex classification system, wheelchair fencing had not been attracting much attention in the sports medicine arena. Recently, the Paralympic Medical Committee had reported the results of the large-scale injury surveillance during the 2012 London Paralympic Games. The injury incidence in wheelchair fencers was reported to be the highest amongst the various wheelchair sporting events during the Paralympic Games. Despite the injury rate of wheelchair fencing is preliminary reported, systematic injury survey for better understanding the injury profile and risks for wheelchair fencing are still lacking. In order to develop an injury prevention strategy that is specific to wheelchair fencing, it is vital to firstly document the prevalence and severity of injury. With reference to the van Mechelen's model of sports injury prevention, further analysis to the identified risks from the injury surveillance would be essential to establish the aetiology and injury mechanism in terms of their biomechanical motions. To date, biomechanical analyses in forms of kinematic and electromyographic analyses are widely used to study the motion in sports science and sports medicine. However, the applications of these studies are new to wheelchair fencing and the precision and accuracy had no figure to reference. Therefore, the reliability and validity of the kinematic and EMG protocol for wheelchair fencing is essential to be established prior to motion tests. By examining and comparing the wheelchair fencing motion characteristics between Category A and B fencers, possible injury mechanism was explored and the potential injury preventive measures were suggested. Therefore, the objectives of the four inter-related studies included in the present thesis were as followings:

Study 1: To quantify and compare the injury pattern between the elite able-bodied and wheelchair foil fencers; and between Category A and Category B fencers

Study 2: To examine the within-session repeatability of three-dimensional kinematics and surface electromyographic measurements during lunge attack motion

Study 3: To examine the validity of the optical tracking methods for rapid upper limb kinematic measurement

Study 4: To quantify and compare the three-dimensional upper limb kinematics and motor recruitment characteristics between Category A and B fencers

7.1.1 Study 1: Musculoskeletal injuries in elite able-bodied and wheelchair foil fencers

The 3-year prospective epidemiology study revealed that wheelchair fencers had higher overall injury incidence rate (3.85/1000 hours) than the able-bodied counterparts (injury incidence rate = 2.41/1000 hours). Upper extremity injuries were predominant in wheelchair fencers (73.8%), with elbow (32.6%) and shoulder strain (15.8%) being the most common injuries. As expected, lower extremity injuries were predominant in able-bodied fencers (69.4%), with muscle strain over knee and thigh region (22.6%), ankle sprain (14.5%), and knee sprain (11.3%) being the leading injuries. For injury risk, wheelchair fencers had higher risk than able-bodied fencers in sustaining minor injury (RR = 2.35; 95% CI, 1.56-3.61), muscle strain (RR = 2.16; 95% CI, 1.34-3.56), shoulder injury (RR = 13.55; 95% CI, 3.39-17.76), and elbow injury (RR = 5.90; 95% CI, 2.45-17.21). When comparing the injury statistics between the two wheelchair fencing groups, fencers with poor trunk control (Category B) were more vulnerable to injuries (4.87/1000 hours) than those with good trunk control (Category A) (2.99/1000 hours). A higher risk in muscle strain (RR = 1.83; 95% CI, 1.04-3.28) and shoulder injury (RR = 4.97; 95% CI, 1.82-16.87) was experienced by Category B fencers as compared to Category A fencers. Results of this study highlighted the distinct injury pattern between the able-bodied and wheelchair fencer groups. Wheelchair fencers without active trunk control (i.e. Category B) have higher risk to sustain from muscle strain and shoulder injury as compared to wheelchair fencers with good trunk ability (i.e. Category A).

The findings of the epidemiology study are thought provoking. It is because it indicates that wheelchair fencers experience higher risks of upper limb injuries while no research has yet been conducted to explore the causes or mechanisms of their upper limb injuries. As such, the kinematic and kinetic experiments in this thesis were the next step to improve our understanding of the biomechanics of wheelchair fencing.

7.1.2 Study 2: Repeatability of kinematic and electromyographic data during lunge attack in wheelchair fencing

The establishment of repeatable kinematic and EMG outcomes is paramount for objective measurement and biomechanical comparison between the Category A and B fencers. This study demonstrated that the current three-dimensional optical motion system and surface EMG measurements during lunge attack motion were highly reliable. Mean $ICC_{3,1}$ and CMC values for angular displacement ranged from 0.73-0.98 and 0.70-0.98 respectively. The mean CMC of the EMG variables of the eight fencing arm muscles ranged from 0.70-0.94; whereas the ICCs of the peak EMG and iEMG values ranged from 0.62-0.93 and 0.72-0.98 respectively. The high CMC and ICC estimates indicated that the optical kinematic method and surface EMG measurements could be validly used for quantifying the upper limb kinematic and electromyographic characteristics of wheelchair fencing.

7.1.3 Study 3: Validity of the optical tracking method for fast upper limb kinematic measurement

The validity of the optical method for 3D upper limb kinematic measurements during rapid movement was established by comparing against the current bench mark - inertial tracking system. The Vicon Motion Analysis System and the Xsens MTx sensors simultaneously captured the six upper limb motions. Pearson's correlation coefficients for shoulder, elbow and wrist joint measurements ranged from 0.71 to 0.87, 0.78 to 0.99, and 0.76 to 0.84, respectively ($p<0.01$). Joint angles as measured by the two systems were within the 95% limits of agreement. Three upper limb motions displayed excellent CMC: shoulder flexion ($r=0.92$), shoulder abduction ($r=0.91$) and elbow flexion ($r=0.95$), $p<0.01$. Optical and inertial tracking systems demonstrated high measurement agreement and correlation for rapid upper limb motion analysis, which implies that both systems are interchangeable for monitoring the kinematics of rapid arm movements (e.g. wheelchair fencing). The optical tracking system is preferable for wheelchair fencing kinematic analysis and used in this thesis.

7.1.4 Study 4: Kinematic and EMG analysis of fencing lunge attack in world-class foil wheelchair fencers

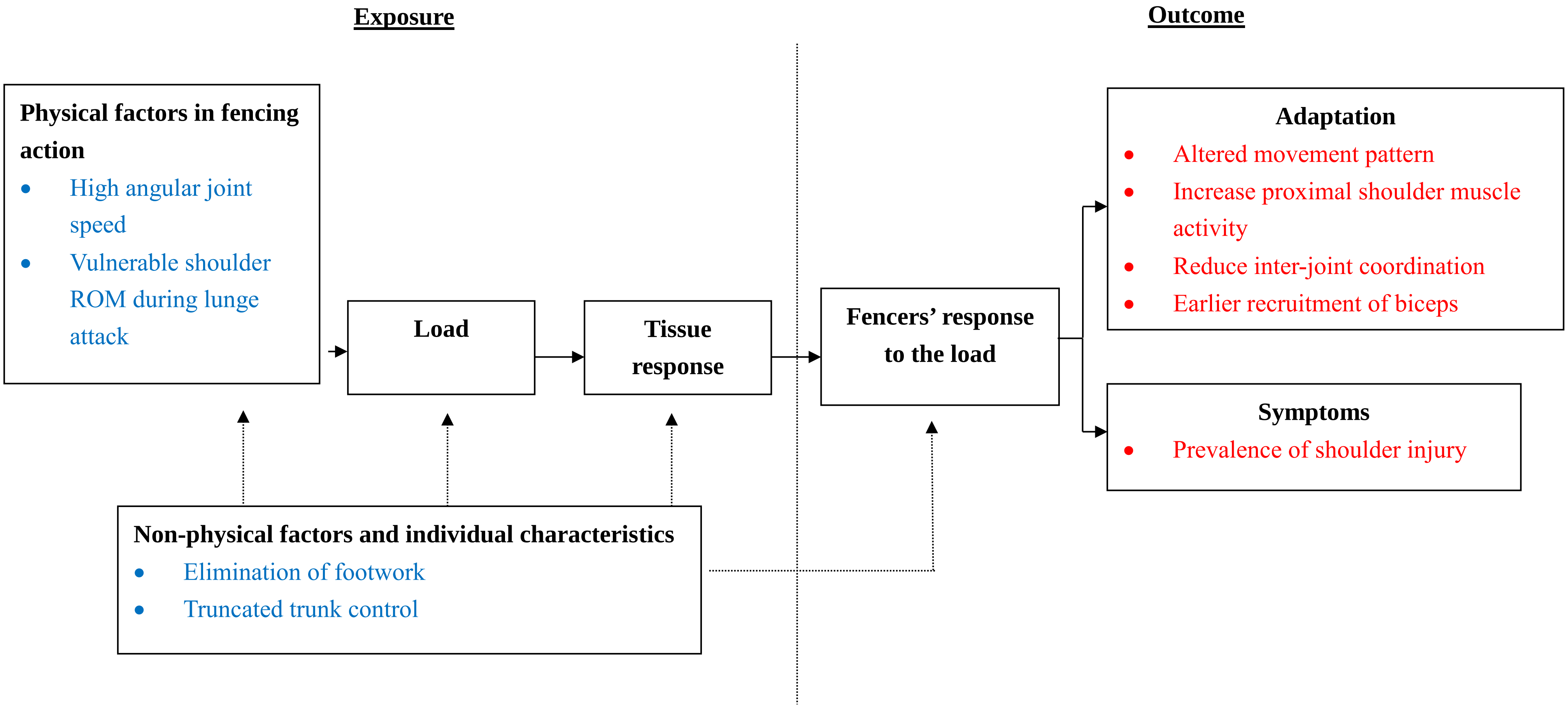
Understanding the upper limb kinematics and electromyographic features of wheelchair fencers with- (Category A) or without-active trunk control (Category B) during lunge attacks provides fundamental information to explore the possible mechanisms of wheelchair fencing injuries. This study documented the lunge attack motion in wheelchair fencing in two ways. First, wheelchair fencers execute a typical powerful lunge attacks by rapidly flexing and abducting their shoulder to 100° to 120° in combination with 50° to 70° shoulder internal rotation. This combined movement may place extra stress to sub-acromial tissues. Repeated practice of this technique may predispose wheelchair fencers to various shoulder problems such as impingement syndrome or rotator cuff disorders that were commonly documented in our previous injury surveillance study. Second, SEMG showed wheelchair fencers from the two groups demonstrated a similar fencing arm muscle recruitment pattern in a proximal-to-distal sequence at all fencing distances; with shoulder muscles (upper trapezius, infraspinatus, anterior-deltoid and mid-deltoid) initiated the lunge, biceps activated eccentrically to stabilize the elbow, triceps were followed to extend the elbow and to advance the weapon towards the target. Wrist flexors and extensors were co-activated to adjust/stabilize the wrist as the foil approached the target.

However, the results of this study also highlighted the divergent upper limb kinematics and EMG between Category A and B fencers during lunge attacks at the longer fencing distances. Three important findings are concluded: i) Compared to the Category A fencers, the Category B fencers demonstrated a longer lunge duration, larger shoulder angular displacement, lower linear and angular velocity at longer fencing length. This discrepancy may imply a specific movement strategy adopted by the Category B fencers to compensate for their poor trunk control. This movement adaptation may induce an excessive stress to the Category B fencers' shoulders and lead to higher incidence of shoulder disorders in this fencer group. ii) A significant decrease of the inter-joint coordination between shoulder and elbow was shown in Category B fencer at longer fencing distances, indicating the reduced coupling between elbow extension and shoulder movements. The dissociated inter-joint coordination may cause a disturbance of the kinetic chain and increases the chance of upper limb injury in category B fencers. iii) Category B fencers are found to have elevated shoulder muscle activities than Category A and have different motor activation patterns over their fencing arm as lunge attack distance increased, potential imposing higher load to the shoulder.

New information obtained from this study provides insight for better understanding the injury mechanism in wheelchair fencing. The shoulder movement during lunge attack has intrinsic risk to increase stress to shoulder subacromial structures that may link with the high incidence of shoulder impingement and rotator cuff disorders found in our previous injury surveillance. The observed change in kinematic pattern and EMG activity in Category B fencers at longer fencing length may also explain the higher risk of this group of fencers to sustain from more shoulder disorders as compared to Category A fencers.

Overall, the present thesis demonstrated that wheelchair fencers have distinct injury patterns. Upper limb injuries are prominent in wheelchair fencers. Category B fencers have higher risk of shoulder injury and muscle strain than Category A fencers. Typical lunge attack in wheelchair fencing follows the proximal-to-distal motor recruitment sequence. Category B fencers displayed a slightly different upper limb biomechanics, with significantly lower the peak horizontal and angular velocities, larger angular displacement and altered joint coordination over their shoulder and elbow joints at longer fencing distances; representing a unique adaptive shoulder movement strategy used by Category B fencers to compensate for their poor trunk control when they maneuver at longer distances. As fencing distance increases, an escalated muscular effort over the shoulder region would be required to remedial the sitting balance of the Category B fencers. When postural demand increases, such adaptive motor strategy may influence the contractility of the fencing arm muscles. The altered kinematics and increased muscular demand, in addition to the highly repetitive nature of wheelchair fencing, may put extra stress and predispose the Category B fencers to have high risk of shoulder disorders (Figure 7.1). However, as the present research design is cross-sectional in nature, the correlation between the alternation of kinematic and/or EMG parameters and shoulder pain in Category B fencers has yet been established in future studies.

Figure 7.1 Possible risks for wheelchair fencing musculoskeletal injuries



7.2 Clinical implications

This study has characterized the injury profiles of wheelchair fencers and the biomechanical difference between wheelchair fencers with good- and poor-trunk control. The preliminary findings render different messages to various personnel of athletes, coaches, sport scientists and medical professionals. Essentially, they should all be aware of the higher risk and unique nature of sports injury in this specific group of disabled athletes. Wheelchair fencers, especially Category B fencers, should be informed about their relatively higher risk in developing shoulder disorders during the active participation in wheelchair fencing. Special attention for early medical consultation and relevant treatment may reduce the deterioration and recurrence of their injuries, and minimize the impairment of daily function.

Coaches for wheelchair fencing, who are commonly proficient in technical knowledge and skills for able-bodied fencing, should recognize the difference in fencing techniques and biomechanics between able-bodied and wheelchair fencing. Specific training program should be designed to optimize individual fencer's performance.

For medical professional, including team physiotherapist, may design appropriate exercise regime to strengthen the key muscles (i.e. the proximal shoulder muscles) for lunge attack so as to prevent injury and rehabilitating those injured athletes. Proper warm-up program or routine training with the emphasis of shoulder mobility may also be useful given the large shoulder angular displacement during lunge attack in longer fencing distance. Equally important, structural evaluation program should be conducted to evaluate the effectiveness of these injury prevention programs.

Our findings provided preliminary evidences that able-bodied training regime and injury prevention strategy should not be directly applied to disabled athletes. Comprehensive injury surveillance, supported by quantitative biomechanical analysis of the disabled sports event should be conducted to investigate the causes of the injury mechanism. The present experimental protocols and findings may serve as the first step to higher quality research in wheelchair fencing in future.

7.3 Limitations of the studies

7.3.1 Subject selection

As an exploratory work, this study included only elite wheelchair fencers with Category A and B classification to minimize the effect of poor skill. Category C was not involved due to the small number of participants that had received the permanent classification status in both Hong Kong and China Paralympic teams. Also, only male subjects were recruited to minimize the potential kinematic and EMG difference due to gender. With all these factors, the findings of this study may not be generalizable to other groups, including novice level, female fencers or wheelchair fencers who belonged to Category C.

7.3.2 Possible contribution of non-fencing arm, trunk and residual lower limb muscles

The upper limb kinematics involved in this study was a combination of trunk and arm motions. The disparity between the two fencing groups could be attributed to the difference in sitting stability control or the non-fencing arm kinetics. This study design did not address or quantify the extent of this combination in affecting the lunge attack. The exact extent of the residual motor control of the hip muscles in the lower limb amputee, for example, or the non-fencing arm assisted in the sitting balance during lunge action were unknown. Although we did not separate the contribution from each of these components, our study did undertake some essential measures. First, the fencers' sitting posture was strictly governed during the tests to minimize the lower limb contribution that might confound the kinematic results. Second, there was no participant reported to have any neurology or injury over their non-fencing arms. It is reasonable to assume that the contribution of the non-fencing arm is similar between the two fencing groups. Future studies on the function of trunk and non-fencing arm during lunge attack in wheelchair fencers are warranted.

7.3.3 Unknown relationship between kinematic / EMG and injury

Although this thesis has identified some difference in injury patterns between able-bodied fencers and wheelchair fencers as well as the biomechanical differences between the two categories of wheelchair fencers, our findings did not provide enough information to formulate the exact mechanism of injury in wheelchair fencers. Injury mechanism in sports is highly complex and interlinked with multiple factors. The situation is further complicated by the distinct intrinsic and extrinsic factors that are discrete to disabled athletes. Given that cognitive, psychosocial, emotional and physical status of the wheelchair fencers may also affect the progress or recovery of injury, future studies are needed to quantify the relative contributions of various factors in relation to the injury pattern in wheelchair fencers.

7.3.4 EMG measurements

Deeply-seated muscle signals could not be detected by surface EMG. Our kinematic data detected shoulder internal rotation despite surface EMG showed strong concurrent activation of the infraspinatus (a shoulder external rotator). The surface EMG could not be detected if the deeper shoulder internal rotators (i.e. subscapularis and pectoralis major) had contributed to the internal rotation of the glenohumeral joint. Other EMG techniques including needle EMG method may be used to locate the deeply-seated muscles. More advanced EMG techniques such as wavelet analysis may also be considered in future.

7.4 Suggestions for further study

7.4.1 Larger scale prospective injury surveillance and biomechanical research

To account for the wide variability of pathologies, diverse extent of disabilities, different competition levels and the complex functional classification in wheelchair fencing, multinational research project including larger sample size of wheelchair fencers are indicated. It is also necessary to have consensus over the definition of injury. Apart from identifying injury based on the types, other definitions (such as duration of absence from training and competitions) should also be considered. For example, more precise computation of training time should be recorded to improve the accuracy on calculating the incidence of injuries, pathology and re-injuries. Furthermore, studies with prospective design are needed to broaden the proposed multi-dimensional injury model of wheelchair fencing.

7.4.2 Investigation for proximal trunk control, non-fencing arm and residual lower limb muscles

Superficial recoding EMG study on other proximal muscle such as latissimus dorsi, pectoralis and serratus would have been of great interest as they may contribute to postural stability at longer reaching length. Investigation of the deeply-located muscles such as the shoulder rotators would provide information for shoulder motor pattern during wheelchair fencing. Besides, further research for non-fencing arm muscles as well as the residual lower limb muscles (in Category A fencers) should be examined to comprehend their level of activation during wheelchair fencing.

7.4.3 Joint force analysis for lunge attack

Future studies should focus on other upper limb kinetics (e.g. the joint force and moment or impact force) during lunge attack in order to provide a more comprehensive analysis of the upper limb mechanical loads experienced by wheelchair fencers. Additional research attempting to predict or simulate the upper limb mechanical loads during lunge attack from upper limb EMG data recorded, with the help of detailed arm-shoulder models, should thereby be encouraged.

7.4.4 Fatigue profile for distal arm muscles

Elbow muscle injury was identified as one of the most common musculoskeletal problems in wheelchair fencers in our injury surveillance. Yet the EMG results revealed the low muscle activities during the lunge attack. Our biomechanical testing has not acquired sufficient information to bridge the gap. Since elbow muscle injury is commonly believed to be related to micro-trauma by the repetitive motions, it is worthwhile to investigate the change of muscle activation level and tendency of fatigue of the forearm muscles during prolonged wheelchair fencing practice in future study.

7.4.5 Other fencing skills and weapons

The experiments in this thesis only examined the lunge attack as it is the most common and important technique for scoring in wheelchair fencing. Since wheelchair fencing is an open-skilled sport and involves numerous fencing techniques, biomechanical research on other techniques and/or fencing weapons are warranted.

Appendix I

The Hong Kong Polytechnic University
Department of Rehabilitation Sciences

Research Project Informed Consent Form

Project title: Three-year prospective injury surveillance study of the Hong Kong elite able-bodied and disabled foil fencers

Investigators: Man CHUNG; PhD student, Department of Rehabilitation Science, Hong Kong Polytechnic University

Supervisors: Dr. Simon YEUNG SS; PhD; Associate Professor, Department of Rehabilitation Science, Hong Kong Polytechnic University

Purpose, value and details of study:

The purpose of this study is to investigate the injury characteristics and possible risk factors of fencing injuries. Results of current research project will establish a foundation for future development of injury prevention and training program for both able-bodied and disabled fencers.

Participants who join this study are on voluntary basis and there will be no personal benefit of any form by joining this study. However, in the case that participants would like to know the details of his/her injuries, the physiotherapy coordinator of this project would provide thorough explanation and relevant advice for managing the identified injuries.

In the coming three years, all participants will be interviewed every two months, with each interview session last for approximately 30 minutes, by the physiotherapists for capturing data of training duration, match duration and injuries (including the types, location and severity of injuries). Injuries to be charted will confine to sports-related only (i.e. during practice session or competition) and the injury should be severe enough to cause at least 1 day absence from practice or competition. All reported injuries will be evaluated by the physiotherapist coordinator of the research project or the chief sports physicians of the Hong Kong Sports Institute Medical Center to provide the diagnosis of injury.

Confidentiality:

All personal and injury data will be kept strictly confidential for the present research use only. Individual information will not be disclosed to any form of reports, journal or presentation before getting the prior approval from individual participant.

Consent:

I, ___________________________, have been explained the details of this study. I voluntarily consent to participate in this study. I understand that I can withdraw from this study at any time without giving reasons, and my withdrawal will not lead to any punishment or prejudice against me. I am aware of any potential risk in joining this study. I also understand that my personal information will not be disclosed to people who are not related to this study and my name or photograph will not appear on any publications resulted from this study.

I can contact the chief investigator, Mr. Man CHUNG at telephone 9659 for any questions about this study. If I have complaints related to the investigator(s), I can contact Ms Michelle Leung, secretary of Departmental Research Committee, at 2766 5397. I know I will be given a signed copy of this consent form.

Signature (subject): Date:

Signature (witness): Date:

Appendix II

香港理工大學康復治療科學系

參與
三年前瞻性香港精英劍擊運動員與傷殘劍擊運動員傷患監管
研究同意書

本人____________________自願及義務參與由香港理工大學博士生鍾惠文先生、香港理工大學康復治療科學系副教授楊世模博士與及英國倫敦 Roehampton University 李潤華教授負責之上述研究。

此項研究的目的是調查劍擊傷害的特性與可能的受傷風險。這個研究的結果希望可以為傷殘劍擊運動員或其他傷殘運動員提供預防運動傷害的方法與有效的訓練方式。於未來三年，參加者將會每隔兩個月接受物理治療師訪問一次，每次訪問為期大約三十分鐘，內容是關於訓練時間、比賽時間與受傷情況（包括類型、位置及程度等）。研究只包括與運動有關的受傷（訓練或比賽），受傷的程度必需達到令參與者缺席練習或比賽一天才會計算在內。參與者的傷勢將會由有關的物理治療師或香港體育學院醫療中心的醫生作出評估或診斷。

研究所得到的個人資料只會用於本研究之中。如未得到參加者的同意，研究所得資料不會向外披露，當中包括以報告、文獻或演說形式。參與者亦需明白其參與純屬義務性質，參與者將不會得到任何形式的利益，故即使參與者欲拒絕繼續參與有關研究，亦不需付上任何責任或導致任何利益上之損失。

本人在此項研究過程中，可提出任何有關研究程序的疑問，並且已應該得到上述之調查負責人員的回應和解釋。假使上述調查之負責人員未能對本人之提問給予滿意的答覆，本人可就有關查詢致電 9659 聯絡鍾惠文先生。若本人對這項研究有任何不滿，亦可以致電 2766 5397 聯絡康復治療科學系研究委員會秘書梁小姐。同時，本人亦知悉此研究的結果，除可能作綜合報告外，本人之個人資料將會保密。本人已經閱畢及完全明白此同意書之內容，並已收到此同意書的副本乙份以作參考。

____________________	____________________
參加者	見證人
____________________	____________________
日期	日期

Appendix III

Training reporting form **CONFIDENTIAL**

Name of athletes: ____________________

Date of first assessment: ____________________

Classification: ____________________

Training record:

Date	Venue	Activities	Duration of training/competition (hours)	Injury	Remark of injury
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	
		Training/competition		Yes / No	

Appendix IV

CONFIDENTIAL

Training / Competition / Injury record for HK fencing delegates

Personal data

Name: ________________________ Gender: ☐ male ☐ female

Age: ________________________ Weapon: ☐ Foil ☐ Epee ☐ Saber

Ranking: ________________________ Years in fencing: ____________________

Fencing arm: ☐ Left ☐ Right

Previous history of injury

Item	Body part	Type of injury	Affecting period	Remark

Training / competition record:

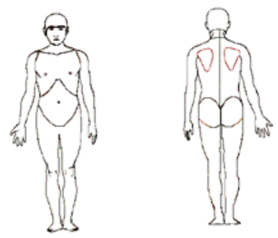
Date	Activities	Number of hours	Injury	Injury code
Jan, 07	Training		Yes / No	
	Competition		Yes / No	
Feb., 07	Training		Yes / No	
	Competition		Yes / No	
Mar., 07	Training		Yes / No	
	Competition		Yes / No	
Apr., 07	Training		Yes / No	
	Competition		Yes / No	
May., 07	Training		Yes / No	
	Competition		Yes / No	
Jun., 07	Training		Yes / No	
	Competition		Yes / No	
Jul., 07	Training		Yes / No	
	Competition		Yes / No	
Aug, 07	Training		Yes / No	
	Competition		Yes / No	
Sept., 07	Training		Yes / No	
	Competition		Yes / No	

Injury charting form for wheelchair fencer

Multiple injuries could be input into one single reporting form

Incidence of injury:	☐ training	☐ match
Type of injury:	☐ new injury	☐ old injury
Causes of injury:	☐ overuse	☐ trauma

Injured body part

☐ head / face	☐ shoulder / clavicle	☐ hip / groin	
☐ neck / cervical	☐ upper arm	☐ thigh	
☐ sternum / ribs / thoracic	☐ elbow	☐ knee	
☐ abdomen	☐ forearm	☐ lower leg	
☐ lumbar / sacrum / pelvis	☐ wrist	☐ ankle	
☐ Non-specific	☐ hand / finger / thumb	☐ foot / toe	

Type of injury

☐ fracture	☐ sprain / ligament injury	☐ contusion / haematoma
☐ other bone injury	☐ muscle rupture / strain / tear	☐ abrasion
☐ dislocation / subluxation	☐ muscle cramp	☐ laceration
☐ meniscus or cartilage injury	☐ tendon injury / rupture / tendonitis	☐ blister
☐ nerve problems	☐ bursitis	☐ fatigue

☐ Other injury (please specify): __ ________________________

Severity of medical illness and injury

☐ minor: 1-7 days loss	☐ moderate: 8-28 days loss	☐ major: > 28 days

Recorder: ______________________ Date: ________________________

Glossary:

Injury:	Any sports injury that causes absence from participation
New injury:	An injury that have never been sustained before
Old injury:	An injury occurring after an initial injury of same type and location
Trauma:	Onset of injury with known history of impact
Overuse:	Injury with insidious onset and without any known trauma.
Fracture:	Traumatic break of bone
Sprain:	Distraction injury of ligaments or joint capsules
Strain:	Distraction injury of muscle and tendons
Contusion:	Tissue bruise without concomitant injuries classified elsewhere
Dislocation:	Partial or complete displacement of bony parts of a joint

Appendix V

The Hong Kong Polytechnic University
Department of Rehabilitation Sciences

Research Project Informed Consent Form

Project title: Reliability of upper limb three-dimensional kinematics and electromyographic analysis during fencing lunge action

Investigators: Man CHUNG; PhD student, Department of Rehabilitation Science, Hong Kong Polytechnic University

Supervisors: Dr. Simon YEUNG SS; PhD; Associate Professor, Department of Rehabilitation Science, Hong Kong Polytechnic University

Professor Raymond LEE; PhD; Professor of Biomechanics and Head of Sport Science, Roehampton University, London

Details of study:

The aim of this study is to evaluate the reliability of upper limb three-dimensional kinematics and electromyographic (EMG) systems during fencing lunge action**.** The results could help to substantiate the accuracy and repeatability of the Vicon motion analysis and EMG systems that will be adopted in future fencing action analysis.

Participants who join this study are on voluntary basis and there will be no personal benefit of any form by joining this study. Participants are required to attend one session of testing that last for approximately 1 hour.

In this study, participants are requested to perform some basic fencing techniques that are performed in maximal speed and effort. At the same time, reflective markers and electrodes will be attached to specific points of the participant's body for motion capture. The joint angles, speed and muscle activities of the tested upper limb will be measured for further reliability analysis.

Dangers and Right:

There is no known risk involved except for possible short-term muscle soreness over the tested upper extremity or low risk of skin allergy due to the reflective markers or electrode placement.

Consent:

I, ___________________________, have been explained the details of this study. I voluntarily consent to participate in this study. I understand that I can withdraw from this study at any time without giving reasons, and my withdrawal will not lead to any punishment or prejudice against me. I am aware of any potential risk in joining this study. I also understand that my personal information will not be disclosed to people who are not related to this study and my name or photograph will not appear on any publications resulted from this study.

I can contact the chief investigator, Mr. Man CHUNG at telephone 9659 for any questions about this study. If I have complaints related to the investigator(s), I can contact Ms Michelle Leung, secretary of Departmental Research Committee, at 2766 5397. I know I will be given a signed copy of this consent form.

Signature (subject): Date:

Signature (witness): Date:

Appendix VI

香港理工大學康復治療科學系

參與
運動姿勢分析系統可靠性測試
研究同意書

本人____________________自願及義務參與由香港理工大學博士生鍾惠文先生、香港理工大學康復治療科學系副教授楊世模博士與及英國倫敦 Roehampton University 李潤華教授負責之上述研究。

參加者於此研究項目中需重覆作出兩至三種指定的基本劍擊動作。參加者身上將會貼上反光標記及電極來協助系統收集高速動態及肌肉活動數據。上肢關節的角度、速度及肌肉活動將會用作分析。所有參加者均需接受一次測試，而測試的時間大約為一小時。此項研究的目的是評估運動姿勢分析裝置對上肢劍擊突刺動作的可靠性，所得的結果將有助往後劍擊動態學的應用及分析。

根據研究人員的專業知識，這些測試除了可能導致肌肉疲勞外，不會為你帶來任何不良的後果。而將貼在參加者身上的光標及電極亦不會引起嚴重的皮膚刺激或過敏反應。此外，參與者亦需明白其參與純屬義務性質，參與者將不會得到任何形式的利益，故即使參與者欲拒絕繼續參與有關研究，亦不需付上任何責任或導致任何利益上之損失。

本人在此項研究過程中，可提出任何有關研究程序的疑問，並且已應該得到上述之調查負責人員的回應和解釋。假使上述調查之負責人員未能對本人之提問給予滿意的答覆，本人可就有關查詢致電 9659 聯絡鍾惠文先生。若本人對這項研究有任何不滿，亦可以致電 27665397 聯絡康復治療科學系研究委員會秘書梁小姐。同時，本人亦知悉此研究的結果，除可能作綜合報告外，本人之個人資料將會保密。本人已經閱畢及完全明白此同意書之內容，並已收到此同意書的副本乙份以作參考。

______________________	______________________
參加者	見證人
______________________	______________________
日期	日期

Appendix VII

MatLab Code for kinematic data processing

```
% setup directory for easy change
direc = 'D:\manchung\Kinematics\100_LL';
files = dir([direc '*.csv']);
samplerate = 100;
condition = input('enter the current condition, eg. 115_XX>', 's');
% read csv trials (n=5)
t1=readtext(fullfile(direc,'100_LL_1.csv'));
t2=readtext(fullfile(direc,'100_LL_2.csv'));
t3=readtext(fullfile(direc,'100_LL_3.csv'));
t4=readtext(fullfile(direc,'100_LL_4.csv'));
t5=readtext(fullfile(direc,'100_LL_5.csv'));
% extract should abduction and data type conversion
sabd1 = t1(12:end,102);
sabd1 = cell2mat(sabd1);
sabd2 = t2(12:end,102);
sabd2 = cell2mat(sabd2);
sabd3 = t3(12:end,102);
sabd3 = cell2mat(sabd3);
sabd4 = t4(12:end,102);
sabd4 = cell2mat(sabd4);
sabd5 = t5(12:end,102);
sabd5 = cell2mat(sabd5);
% % extract should flexion and data type conversion
sflex1 = t1(12:end,104);
sflex1 = cell2mat(sflex1);
sflex2 = t2(12:end,104);
sflex2 = cell2mat(sflex2);
sflex3 = t3(12:end,104);
sflex3 = cell2mat(sflex3);
sflex4 = t4(12:end,104);
sflex4 = cell2mat(sflex4);
sflex5 = t5(12:end,104);
sflex5 = cell2mat(sflex5);
% % extract should rotation and data type conversion
srot1 = t1(12:end,106);
```

```
srot1 = cell2mat(srot1);
srot2 = t2(12:end,106);
srot2 = cell2mat(srot2);
srot3 = t3(12:end,106);
srot3 = cell2mat(srot3);
srot4 = t4(12:end,106);
srot4 = cell2mat(srot4);
srot5 = t5(12:end,106);
srot5 = cell2mat(srot5);
%% extract elbow flexion and data type conversion
eflex1 = t1(12:end,107);
eflex1 = cell2mat(eflex1);
eflex2 = t2(12:end,107);
eflex2 = cell2mat(eflex2);
eflex3 = t3(12:end,107);
eflex3 = cell2mat(eflex3);
eflex4 = t4(12:end,107);
eflex4 = cell2mat(eflex4);
eflex5 = t5(12:end,107);
eflex5 = cell2mat(eflex5);
%% extract forearm supination/pronation and data type conversion
fsp1 = t1(12:end,109);
fsp1 = cell2mat(fsp1);
fsp2 = t2(12:end,109);
fsp2 = cell2mat(fsp2);
fsp3 = t3(12:end,109);
fsp3 = cell2mat(fsp3);
fsp4 = t4(12:end,109);
fsp4 = cell2mat(fsp4);
fsp5 = t5(12:end,109);
fsp5 = cell2mat(fsp5);
%% extract wrist flexion and data type conversion
wflex1 = t1(12:end,110);
wflex1 = cell2mat(wflex1);
wflex2 = t2(12:end,110);
wflex2 = cell2mat(wflex2);
wflex3 = t3(12:end,110);
wflex3 = cell2mat(wflex3);
```

```
wflex4 = t4(12:end,110);
wflex4 = cell2mat(wflex4);
wflex5 = t5(12:end,110);
wflex5 = cell2mat(wflex5);
% % extract wrist radial/ulnar deviation and data type conversion
wru1 = t1(12:end,111);
wru1 = cell2mat(wru1);
wru2 = t2(12:end,111);
wru2 = cell2mat(wru2);
wru3 = t3(12:end,111);
wru3 = cell2mat(wru3);
wru4 = t4(12:end,111);
wru4 = cell2mat(wru4);
wru5 = t5(12:end,111);
wru5 = cell2mat(wru5);
 % Horizontal velocity of shoulder, elbow, and wrist
hor_shY1 = gradient((cell2mat(t1(12:end,15)))*samplerate);
hor_shY2 = gradient((cell2mat(t2(12:end,15)))*samplerate);
hor_shY3 = gradient((cell2mat(t3(12:end,15)))*samplerate);
hor_shY4 = gradient((cell2mat(t4(12:end,15)))*samplerate);
hor_shY5 = gradient((cell2mat(t5(12:end,15)))*samplerate);
mhor_shY = [max(hor_shY1) max(hor_shY2) max(hor_shY3) max(hor_shY4) max(hor_shY5)];
mhor_shY = [min(hor_shY1) min(hor_shY2) min(hor_shY3) min(hor_shY4) min(hor_shY5)];
mhor_shY = [mean(hor_shY1) mean(hor_shY2) mean(hor_shY3) mean(hor_shY4) mean(hor_shY5)];
mhor_shY_max = max(mhor_shY);
mhor_shY_min = min(mhor_shY);
mhor_shY_mean = mean(mhor_shY);
hor_elbY1 = gradient((cell2mat(t1(12:end,27)))*samplerate);
hor_elbY2 = gradient((cell2mat(t2(12:end,27)))*samplerate);
hor_elbY3 = gradient((cell2mat(t3(12:end,27)))*samplerate);
hor_elbY4 = gradient((cell2mat(t4(12:end,27)))*samplerate);
hor_elbY5 = gradient((cell2mat(t5(12:end,27)))*samplerate);
mhor_elbY = [max(hor_elbY1) max(hor_elbY2) max(hor_elbY3) max(hor_elbY4) max(hor_elbY5)];
mhor_elbY = [min(hor_elbY1) min(hor_elbY2) min(hor_elbY3) min(hor_elbY4) min(hor_elbY5)];
mhor_elbY = [mean(hor_elbY1) mean(hor_elbY2) mean(hor_elbY3) mean(hor_elbY4) mean(hor_elbY5)];
mhor_elbY_max = max(mhor_elbY);
mhor_elbY_min = min(mhor_elbY);
mhor_elbY_mean = mean(mhor_elbY);
```

```
hor_wrY1 = gradient((cell2mat(t1(12:end,33)))*samplerate);
hor_wrY2 = gradient((cell2mat(t2(12:end,33)))*samplerate);
hor_wrY3 = gradient((cell2mat(t3(12:end,33)))*samplerate);
hor_wrY4 = gradient((cell2mat(t4(12:end,33)))*samplerate);
hor_wrY5 = gradient((cell2mat(t5(12:end,33)))*samplerate);
mhor_wrY = [max(hor_wrY1) max(hor_wrY2) max(hor_wrY3) max(hor_wrY4) max(hor_wrY5)];
mhor_wrY = [min(hor_wrY1) min(hor_wrY2) min(hor_wrY3) min(hor_wrY4) min(hor_wrY5)];
mhor_wrY = [mean(hor_wrY1) mean(hor_wrY2) mean(hor_wrY3) mean(hor_wrY4) mean(hor_wrY5)];
mhor_wrY_max = max(mhor_wrY);
mhor_wrY_min = min(mhor_wrY);
mhor_wrY_mean = mean(mhor_wrY);
% Vertical velocity of shoulder, elbow, and wrist
ver_shY1 = gradient((cell2mat(t1(12:end,16)))*samplerate);
ver_shY2 = gradient((cell2mat(t2(12:end,16)))*samplerate);
ver_shY3 = gradient((cell2mat(t3(12:end,16)))*samplerate);
ver_shY4 = gradient((cell2mat(t4(12:end,16)))*samplerate);
ver_shY5 = gradient((cell2mat(t5(12:end,16)))*samplerate);
mver_shY = [max(ver_shY1) max(ver_shY2) max(ver_shY3) max(ver_shY4) max(ver_shY5)];
mver_shY = [min(ver_shY1) min(ver_shY2) min(ver_shY3) min(ver_shY4) min(ver_shY5)];
mver_shY = [mean(ver_shY1) mean(ver_shY2) mean(ver_shY3) mean(ver_shY4) mean(ver_shY5)];
mver_shY_max = max(mver_shY);
mver_shY_min = min(mver_shY);
mver_shY_mean = mean(mver_shY);
ver_elbY1 = gradient((cell2mat(t1(12:end,28)))*samplerate);
ver_elbY2 = gradient((cell2mat(t2(12:end,28)))*samplerate);
ver_elbY3 = gradient((cell2mat(t3(12:end,28)))*samplerate);
ver_elbY4 = gradient((cell2mat(t4(12:end,28)))*samplerate);
ver_elbY5 = gradient((cell2mat(t5(12:end,28)))*samplerate);
mver_elbY = [max(ver_elbY1) max(ver_elbY2) max(ver_elbY3) max(ver_elbY4) max(ver_elbY5)];
mver_elbY = [min(ver_elbY1) min(ver_elbY2) min(ver_elbY3) min(ver_elbY4) min(ver_elbY5)];
mver_elbY = [mean(ver_elbY1) mean(ver_elbY2) mean(ver_elbY3) mean(ver_elbY4) mean(ver_elbY5)];
mver_elbY_max = max(mver_elbY);
mver_elbY_min = min(mver_elbY);
mver_elbY_mean = mean(mver_elbY);
ver_wrY1 = gradient((cell2mat(t1(12:end,34)))*samplerate);
ver_wrY2 = gradient((cell2mat(t2(12:end,34)))*samplerate);
ver_wrY3 = gradient((cell2mat(t3(12:end,34)))*samplerate);
ver_wrY4 = gradient((cell2mat(t4(12:end,34)))*samplerate);
```

```
ver_wrY5 = gradient((cell2mat(t5(12:end,34)))*samplerate);
mver_wrY = [max(ver_wrY1) max(ver_wrY2) max(ver_wrY3) max(ver_wrY4) max(ver_wrY5)];
mver_wrY = [min(ver_wrY1) min(ver_wrY2) min(ver_wrY3) min(ver_wrY4) min(ver_wrY5)];
mver_wrY = [mean(ver_wrY1) mean(ver_wrY2) mean(ver_wrY3) mean(ver_wrY4) mean(ver_wrY5)];
mver_wrY_max = max(mver_wrY);
mver_wrY_min = min(mver_wrY);
mver_wrY_mean = mean(mver_wrY);
% % normalization and collection of trials
sabd1 = normalizer(sabd1,101,'linear');
sabd2 = normalizer(sabd2,101,'linear');
sabd3 = normalizer(sabd3,101,'linear');
sabd4 = normalizer(sabd4,101,'linear');
sabd5 = normalizer(sabd5,101,'linear');
sabd = [sabd1 sabd2 sabd3 sabd4 sabd5];
sflex1 = normalizer(sflex1,101,'linear');
sflex2 = normalizer(sflex2,101,'linear');
sflex3 = normalizer(sflex3,101,'linear');
sflex4 = normalizer(sflex4,101,'linear');
sflex5 = normalizer(sflex5,101,'linear');
sflex = [sflex1 sflex2 sflex3 sflex4 sflex5];
srot1 = normalizer(srot1,101,'linear');
srot2 = normalizer(srot2,101,'linear');
srot3 = normalizer(srot3,101,'linear');
srot4 = normalizer(srot4,101,'linear');
srot5 = normalizer(srot5,101,'linear');
srot = [srot1 srot2 srot3 srot4 srot5];
eflex1 = normalizer(eflex1,101,'linear');
eflex2 = normalizer(eflex2,101,'linear');
eflex3 = normalizer(eflex3,101,'linear');
eflex4 = normalizer(eflex4,101,'linear');
eflex5 = normalizer(eflex5,101,'linear');
eflex = [eflex1 eflex2 eflex3 eflex4 eflex5];
fsp1 = normalizer(fsp1,101,'linear');
fsp2 = normalizer(fsp2,101,'linear');
fsp3 = normalizer(fsp3,101,'linear');
fsp4 = normalizer(fsp4,101,'linear');
fsp5 = normalizer(fsp5,101,'linear');
fsp = [fsp1 fsp2 fsp3 fsp4 fsp5];
```

```
wflex1 = normalizer(wflex1,101,'linear');
wflex2 = normalizer(wflex2,101,'linear');
wflex3 = normalizer(wflex3,101,'linear');
wflex4 = normalizer(wflex4,101,'linear');
wflex5 = normalizer(wflex5,101,'linear');
wflex = [wflex1 wflex2 wflex3 wflex4 wflex5];
wru1 = normalizer(wru1,101,'linear');
wru2 = normalizer(wru2,101,'linear');
wru3 = normalizer(wru3,101,'linear');
wru4 = normalizer(wru4,101,'linear');
wru5 = normalizer(wru5,101,'linear');
wru = [wru1 wru2 wru3 wru4 wru5];

% Analysis between trials
msabd = mean(sabd,2); msabd_sd = std(sabd,0, 2); msabd_max = max(msabd); msabd_min = min(msabd);
msflex = mean(sflex,2); msflex_sd = std(sflex,0, 2); msflex_max = max(msflex); msflex_min = min(msflex);
msrot = mean(srot,2); msrot_sd = std(srot,0, 2); msrot_max = max(msrot); msrot_min = min(msrot);
meflex = mean(eflex,2); meflex_sd = std(eflex,0, 2); meflex_max = max(meflex); meflex_min = min(meflex);
mfsp = mean(fsp,2); mfsp_sd = std(fsp,0, 2); mfsp_max = max(mfsp); mfsp_min = min(mfsp);
mwflex = mean(wflex,2); mwflex_sd = std(wflex,0, 2); mwflex_max = max(mwflex); mwflex_min = min(mwflex);
mwru = mean(wru,2); mwru_sd = std(wru,0, 2); mwru_max = max(mwru); mwru_min = min(mwru);

% Analysis of velocity
vel_sabd1 = gradient(sabd1);
vel_sabd2 = gradient(sabd2);
vel_sabd3 = gradient(sabd3);
vel_sabd4 = gradient(sabd4);
vel_sabd5 = gradient(sabd5);
vel_sabd = [vel_sabd1 vel_sabd2 vel_sabd3 vel_sabd4 vel_sabd5];
mvel_sabd = mean(vel_sabd, 2); vel_sabd_sd = std(vel_sabd,0, 2);

vel_sflex1 = gradient(sflex1);
vel_sflex2 = gradient(sflex2);
vel_sflex3 = gradient(sflex3);
vel_sflex4 = gradient(sflex4);
vel_sflex5 = gradient(sflex5);
vel_sflex = [vel_sflex1 vel_sflex2 vel_sflex3 vel_sflex4 vel_sflex5];
mvel_sflex = mean(vel_sflex, 2); vel_sflex_sd = std(vel_sflex,0, 2);
```

```
vel_srot1 = gradient(srot1);
vel_srot2 = gradient(srot2);
vel_srot3 = gradient(srot3);
vel_srot4 = gradient(srot4);
vel_srot5 = gradient(srot5);
vel_srot = [vel_srot1 vel_srot2 vel_srot3 vel_srot4 vel_srot5];
mvel_srot =  mean(vel_srot, 2); vel_srot_sd = std(vel_srot,0, 2);

vel_eflex1 = gradient(eflex1);
vel_eflex2 = gradient(eflex2);
vel_eflex3 = gradient(eflex3);
vel_eflex4 = gradient(eflex4);
vel_eflex5 = gradient(eflex5);
vel_eflex = [vel_eflex1 vel_eflex2 vel_eflex3 vel_eflex4 vel_eflex5];
mvel_eflex = mean(vel_eflex, 2); vel_eflex_sd = std(vel_eflex,0, 2);

vel_fsp1 = gradient(fsp1);
vel_fsp2 = gradient(fsp2);
vel_fsp3 = gradient(fsp3);
vel_fsp4 = gradient(fsp4);
vel_fsp5 = gradient(fsp5);
vel_fsp = [vel_fsp1 vel_fsp2 vel_fsp3 vel_fsp4 vel_fsp5];
mvel_fsp = mean(vel_fsp, 2); vel_fsp_sd = std(vel_fsp,0, 2);

vel_wflex1 = gradient(wflex1);
vel_wflex2 = gradient(wflex2);
vel_wflex3 = gradient(wflex3);
vel_wflex4 = gradient(wflex4);
vel_wflex5 = gradient(wflex5);
vel_wflex = [vel_wflex1 vel_wflex2 vel_wflex3 vel_wflex4 vel_wflex5];
mvel_wflex = mean(vel_wflex, 2); vel_wflex_sd = std(vel_wflex,0, 2);

vel_wru1 = gradient(wru1);
vel_wru2 = gradient(wru2);
vel_wru3 = gradient(wru3);
vel_wru4 = gradient(wru4);
vel_wru5 = gradient(wru5);
```

```
vel_wru = [vel_wru1 vel_wru2 vel_wru3 vel_wru4 vel_wru5];
mvel_wru = mean(vel_wru, 2); vel_wru_sd = std(vel_wru,0, 2);


% Analysis of acceleration
% % a_sabd = gradient(vel_sabd); [max_asabd max_asabd_dc] = max(a_sabd); [min_asabd min_asabd_dc] =
min(a_sabd);a_sabd_sd = std(a_sabd);
% % a_sflex = gradient(vel_sflex); [max_asflex max_asflex_dc] = max(a_sflex); [min_asflex min_asflex_dc]
= min(a_sflex);a_sflex_sd = std(a_sflex);
% % a_srot = gradient(vel_srot); [max_asrot max_asrot_dc] = max(a_srot); [min_asrot min_asrot_dc] =
min(a_srot);a_srot_sd = std(a_srot);
% % a_eflex = gradient(vel_eflex); [max_aeflex max_aeflex_dc] = max(a_eflex); [min_aeflex min_aeflex_dc]
= min(a_eflex);a_eflex_sd = std(a_eflex);
% % a_fsp = gradient(vel_fsp); [max_afsp max_afsp_dc] = max(a_fsp); [min_afsp min_afsp_dc] =
min(a_fsp);a_fsp_sd = std(a_fsp);
% % a_wflex = gradient(vel_wflex); [max_awflex max_awflex_dc] = max(a_wflex); [min_awflex min_awflex_dc]
= min(a_wflex);a_wflex_sd = std(a_wflex);
% % a_wru = gradient(vel_wru); [max_awru max_awru_dc] = max(a_wru); [min_awru min_awru_dc] =
min(a_wru);a_wru_sd = std(a_wru);
a_sabd1 = gradient(vel_sabd1);
a_sabd2 = gradient(vel_sabd2);
a_sabd3 = gradient(vel_sabd3);
a_sabd4 = gradient(vel_sabd4);
a_sabd5 = gradient(vel_sabd5);
a_sabd = [a_sabd1 a_sabd2 a_sabd3 a_sabd4 a_sabd5];
ma_sabd = mean(a_sabd, 2); a_sabd_sd = std(a_sabd,0, 2);

a_sflex1 = gradient(vel_sflex1);
a_sflex2 = gradient(vel_sflex2);
a_sflex3 = gradient(vel_sflex3);
a_sflex4 = gradient(vel_sflex4);
a_sflex5 = gradient(vel_sflex5);
a_sflex = [a_sflex1 a_sflex2 a_sflex3 a_sflex4 a_sflex5];
ma_sflex = mean(a_sflex, 2); a_sflex_sd = std(a_sflex,0, 2);

a_srot1 = gradient(vel_srot1);
a_srot2 = gradient(vel_srot2);
a_srot3 = gradient(vel_srot3);
```

```
a_srot4 = gradient(vel_srot4);
a_srot5 = gradient(vel_srot5);
a_srot = [a_srot1 a_srot2 a_srot3 a_srot4 a_srot5];
ma_srot =  mean(a_srot, 2); a_srot_sd = std(a_srot,0, 2);

a_eflex1 = gradient(vel_eflex1);
a_eflex2 = gradient(vel_eflex2);
a_eflex3 = gradient(vel_eflex3);
a_eflex4 = gradient(vel_eflex4);
a_eflex5 = gradient(vel_eflex5);
a_eflex = [a_eflex1 a_eflex2 a_eflex3 a_eflex4 a_eflex5];
ma_eflex = mean(a_eflex, 2); a_eflex_sd = std(a_eflex,0, 2);

a_fsp1 = gradient(vel_fsp1);
a_fsp2 = gradient(vel_fsp2);
a_fsp3 = gradient(vel_fsp3);
a_fsp4 = gradient(vel_fsp4);
a_fsp5 = gradient(vel_fsp5);
a_fsp = [a_fsp1 a_fsp2 a_fsp3 a_fsp4 a_fsp5];
ma_fsp = mean(a_fsp, 2); a_fsp_sd = std(a_fsp,0, 2);

a_wflex1 = gradient(vel_wflex1);
a_wflex2 = gradient(vel_wflex2);
a_wflex3 = gradient(vel_wflex3);
a_wflex4 = gradient(vel_wflex4);
a_wflex5 = gradient(vel_wflex5);
a_wflex = [a_wflex1 a_wflex2 a_wflex3 a_wflex4 a_wflex5];
ma_wflex = mean(a_wflex, 2); a_wflex_sd = std(a_wflex,0, 2);

a_wru1 = gradient(vel_wru1);
a_wru2 = gradient(vel_wru2);
a_wru3 = gradient(vel_wru3);
a_wru4 = gradient(vel_wru4);
a_wru5 = gradient(vel_wru5);
a_wru = [a_wru1 a_wru2 a_wru3 a_wru4 a_wru5];
ma_wru = mean(a_wru, 2); a_wru_sd = std(a_wru,0, 2);

% Plot results
```

```
% figure(1), title('Shoulder abduction 5 trials'); hold on;
% figure(1), plot((sabd), 'k-');
% figure(2), title('Mean shoulder abduction'); hold on;
% figure(2), plot((msabd), 'b-'); hold on;
% figure(2), plot((msabd + msabd_sd), 'b--'); hold on;
% figure(2), plot((msabd - msabd_sd), 'b--');
%
% figure(3), title('Shoulder flexion 5 trials'); hold on;
% figure(3), plot((sflex), 'k-');
% figure(4), title('Mean shoulder flexion'); hold on;
% figure(4), plot((msflex), 'b-'); hold on;
% figure(4), plot((msflex + msflex_sd), 'b--'); hold on;
% figure(4), plot((msflex - msflex_sd), 'b--');
%
% figure(5), title('Shoulder rotation 5 trials'); hold on;
% figure(5), plot((srot), 'k-');
% figure(6), title('Mean shoulder rotation'); hold on;
% figure(6), plot((msrot), 'b-'); hold on;
% figure(6), plot((msrot + msrot_sd), 'b--'); hold on;
% figure(6), plot((msrot - msrot_sd), 'b--');
%
% figure(7), title('Elbow flexion 5 trials'); hold on;
% figure(7), plot((eflex), 'k-');
% figure(8), title('Mean elbow flexion'); hold on;
% figure(8), plot((meflex), 'b-'); hold on;
% figure(8), plot((meflex + meflex_sd), 'b--'); hold on;
% figure(8), plot((meflex - meflex_sd), 'b--');
%
% figure(9), title('Forearm supination/pronation 5 trials'); hold on;
% figure(9), plot((fsp), 'k-');
% figure(10), title('Mean forearm sup/pronation'); hold on;
% figure(10), plot((mfsp), 'b-'); hold on;
% figure(10), plot((mfsp + mfsp_sd), 'b--'); hold on;
% figure(10), plot((mfsp - mfsp_sd), 'b--');
%
% figure(11), title('Wrist flexion 5 trials'); hold on;
% figure(11), plot((wflex), 'k-');
% figure(12), title('Mean wrist flexion'); hold on;
```

```
% figure(12), plot((mwflex), 'b-'); hold on;
% figure(12), plot((mwflex + mwflex_sd), 'b--'); hold on;
% figure(12), plot((mwflex - mwflex_sd), 'b--');
%
% figure(13), title('Wrist radial/ulnar deviation 5 trials'); hold on;
% figure(13), plot((wru), 'k-');
% figure(14), title('Mean wrist RU deviation'); hold on;
% figure(14), plot((mwru), 'b-'); hold on;
% figure(14), plot((mwru + mwru_sd), 'b--'); hold on;
% figure(14), plot((mwru - mwru_sd), 'b--');

angle_7_values= [msabd, msflex, msrot, meflex, mfsp, mwflex, mwru];
ang_vel_7_values= [mvel_sabd, mvel_sflex, mvel_srot, mvel_eflex, mvel_fsp, mvel_wflex, mvel_wru];
ang_acc_7_values= [ma_sabd, ma_sflex, ma_srot, ma_eflex, ma_fsp, ma_wflex, ma_wru];

angle_sd = [msabd_sd, msflex_sd, msrot_sd, meflex_sd, mfsp_sd, mwflex_sd, mwru_sd];
[max_angle_sd max_angle_sd_dc] = max(angle_sd);
[min_angle_sd min_angle_sd_dc] = min(angle_sd);
ang_vel_sd = [vel_sabd_sd, vel_sflex_sd, vel_srot_sd, vel_eflex_sd, vel_fsp_sd, vel_wflex_sd, vel_wru_sd];
[max_ang_vel_sd max_ang_vel_sd_dc] = max(ang_vel_sd);
[min_ang_vel_sd min_ang_vel_sd_dc] = min(ang_vel_sd);
ang_acc_sd = [a_sabd_sd, a_sflex_sd, a_srot_sd, a_eflex_sd, a_fsp_sd, a_wflex_sd, a_wru_sd];
[max_ang_acc_sd max_ang_acc_sd_dc] = max(ang_acc_sd);
[min_ang_acc_sd min_ang_acc_sd_dc] = min(ang_acc_sd);

hori_vel = [mhor_shY_mean, mhor_elbY_mean, mhor_wrY_mean];
vert_vel = [mver_shY_mean, mver_elbY_mean, mver_wrY_mean];




heading_angle = [];
heading_angle{1,1} = 'sabd'
heading_angle{1,2} = 'sflex'
heading_angle{1,3} = 'srot'
heading_angle{1,4} = 'eflex'
heading_angle{1,5} = 'fsp'
heading_angle{1,6} = 'wflex'
```

```
heading_angle{1,7} = 'wru'
heading_vel = [];
heading_vel{1,1} = 'vel_sabd'
heading_vel{1,2} = 'vel_sflex'
heading_vel{1,3} = 'vel_srot'
heading_vel{1,4} = 'vel_eflex'
heading_vel{1,5} = 'vel_fsp'
heading_vel{1,6} = 'vel_wflex'
heading_vel{1,7} = 'vel_wru'
heading_acc = [];
heading_acc{1,1} = 'a_sabd'
heading_acc{1,2} = 'a_sflex'
heading_acc{1,3} = 'a_srot'
heading_acc{1,4} = 'a_eflex'
heading_acc{1,5} = 'a_fsp'
heading_acc{1,6} = 'a_wflex'
heading_acc{1,7} = 'a_wru'
heading_3 = [];
heading_3{1,1} = 'shoulder'
heading_3{1,2} = 'elbow'
heading_3{1,3} = 'wrist'

angle_t1 = [sabd1 sflex1 srot1 eflex1 fsp1 wflex1 wru1];
angle_t2 = [sabd2 sflex2 srot2 eflex2 fsp2 wflex2 wru2];
angle_t3 = [sabd3 sflex3 srot3 eflex3 fsp3 wflex3 wru3];
angle_t4 = [sabd4 sflex4 srot4 eflex4 fsp4 wflex4 wru4];
angle_t5 = [sabd5 sflex5 srot5 eflex5 fsp5 wflex5 wru5];

vel_t1 = [vel_sabd1 vel_sflex1 vel_srot1 vel_eflex1 vel_fsp1 vel_wflex1 vel_wru1];
vel_t2 = [vel_sabd2 vel_sflex2 vel_srot2 vel_eflex2 vel_fsp2 vel_wflex2 vel_wru2];
vel_t3 = [vel_sabd3 vel_sflex3 vel_srot3 vel_eflex3 vel_fsp3 vel_wflex3 vel_wru3];
vel_t4 = [vel_sabd4 vel_sflex4 vel_srot4 vel_eflex4 vel_fsp4 vel_wflex4 vel_wru4];
vel_t5 = [vel_sabd5 vel_sflex5 vel_srot5 vel_eflex5 vel_fsp5 vel_wflex5 vel_wru5];

a_t1 = [a_sabd1 a_sflex1 a_srot1 a_eflex1 a_fsp1 a_wflex1 a_wru1];
a_t2 = [a_sabd2 a_sflex2 a_srot2 a_eflex2 a_fsp2 a_wflex2 a_wru2];
a_t3 = [a_sabd3 a_sflex3 a_srot3 a_eflex3 a_fsp3 a_wflex3 a_wru3];
a_t4 = [a_sabd4 a_sflex4 a_srot4 a_eflex4 a_fsp4 a_wflex4 a_wru4];
```

```
a_t5 = [a_sabd5 a_sflex5 a_srot5 a_eflex5 a_fsp5 a_wflex5 a_wru5];

hori_t1 = [hor_shY1 hor_elbY1 hor_wrY1];
hori_t2 = [hor_shY2 hor_elbY2 hor_wrY2];
hori_t3 = [hor_shY3 hor_elbY3 hor_wrY3];
hori_t4 = [hor_shY4 hor_elbY4 hor_wrY4];
hori_t5 = [hor_shY5 hor_elbY5 hor_wrY5];

ver_t1 = [ver_shY1 ver_elbY1 ver_wrY1];
ver_t2 = [ver_shY2 ver_elbY2 ver_wrY2];
ver_t3 = [ver_shY3 ver_elbY3 ver_wrY3];
ver_t4 = [ver_shY4 ver_elbY4 ver_wrY4];
ver_t5 = [ver_shY5 ver_elbY5 ver_wrY5];

t1 = 't1';
t2 = 't2';
t3 = 't3';
t4 = 't4';
t5 = 't5';

xlswrite(t1, angle_t1, 'angle', 'A2');
xlswrite(t1, heading_angle, 'angle');
xlswrite(t1, vel_t1, 'ang_velocity', 'A2');
xlswrite(t1, heading_vel, 'ang_velocity');
xlswrite(t1, a_t1, 'ang_acc', 'A2');
xlswrite(t1, heading_acc, 'ang_acc');
xlswrite(t1, hori_t1, 'hori_vel', 'A2');
xlswrite(t1, heading_3, 'hori_vel');
xlswrite(t1, ver_t1, 'vert_vel', 'A2');
xlswrite(t1, heading_3, 'vert_vel');
xlswrite(t2, angle_t2, 'angle', 'A2');
xlswrite(t2, heading_angle, 'angle');
xlswrite(t2, vel_t2, 'ang_velocity', 'A2');
xlswrite(t2, heading_vel, 'ang_velocity');
xlswrite(t2, a_t2, 'ang_acc', 'A2');
xlswrite(t2, heading_acc, 'ang_acc');
xlswrite(t2, hori_t2, 'hori_vel', 'A2');
xlswrite(t2, heading_3, 'hori_vel');
```

```
xlswrite(t2, ver_t2, 'vert_vel', 'A2');
xlswrite(t2, heading_3, 'vert_vel');
xlswrite(t3, angle_t3, 'angle', 'A2');
xlswrite(t3, heading_angle, 'angle');
xlswrite(t3, vel_t3, 'ang_velocity', 'A2');
xlswrite(t3, heading_vel, 'ang_velocity');
xlswrite(t3, a_t1, 'ang_acc', 'A2');
xlswrite(t3, heading_acc, 'ang_acc');
xlswrite(t3, hori_t3, 'hori_vel', 'A2');
xlswrite(t3, heading_3, 'hori_vel');
xlswrite(t3, ver_t3, 'vert_vel', 'A2');
xlswrite(t3, heading_3, 'vert_vel');
xlswrite(t4, angle_t4, 'angle', 'A2');
xlswrite(t4, heading_angle, 'angle');
xlswrite(t4, vel_t4, 'ang_velocity', 'A2');
xlswrite(t4, heading_vel, 'ang_velocity');
xlswrite(t4, a_t4, 'ang_acc', 'A2');
xlswrite(t4, heading_acc, 'ang_acc');
xlswrite(t4, hori_t4, 'hori_vel', 'A2');
xlswrite(t4, heading_3, 'hori_vel');
xlswrite(t4, ver_t4, 'vert_vel', 'A2');
xlswrite(t4, heading_3, 'vert_vel');
xlswrite(t5, angle_t5, 'angle', 'A2');
xlswrite(t5, heading_angle, 'angle');
xlswrite(t5, vel_t5, 'ang_velocity', 'A2');
xlswrite(t5, heading_vel, 'ang_velocity');
xlswrite(t5, a_t5, 'ang_acc', 'A2');
xlswrite(t5, heading_acc, 'ang_acc');
xlswrite(t5, hori_t5, 'hori_vel', 'A2');
xlswrite(t5, heading_3, 'hori_vel');
xlswrite(t5, ver_t5, 'vert_vel', 'A2');
xlswrite(t5, heading_3, 'vert_vel');

xlswrite(condition, angle_7_values, 'angle_mean', 'A2');
xlswrite(condition, heading_angle, 'angle_mean');
xlswrite(condition, angle_sd, 'angle_SD', 'A2');
xlswrite(condition, heading_angle, 'angle_SD');
xlswrite(condition, max_angle_sd, 'max_angle_SD', 'A2');
```

```
xlswrite(condition, max_angle_sd_dc, 'max_angle_SD', 'A3');
xlswrite(condition, heading_angle, 'max_angle_SD');
xlswrite(condition, min_angle_sd, 'min_angle_SD', 'A2');
xlswrite(condition, min_angle_sd_dc, 'min_angle_SD', 'A3');
xlswrite(condition, heading_angle, 'min_angle_SD');

xlswrite(condition, ang_vel_7_values, 'ang_vel', 'A2');
xlswrite(condition, heading_vel, 'ang_vel');
xlswrite(condition, ang_vel_sd, 'ang_vel_SD', 'A2');
xlswrite(condition, heading_vel, 'ang_vel_SD');
xlswrite(condition, max_ang_vel_sd, 'max_ang_vel_SD', 'A2');
xlswrite(condition, max_ang_vel_sd_dc, 'max_ang_vel_SD', 'A3');
xlswrite(condition, heading_vel, 'max_ang_vel_SD');
xlswrite(condition, min_ang_vel_sd, 'min_ang_vel_SD', 'A2');
xlswrite(condition, min_ang_vel_sd_dc, 'min_ang_vel_SD', 'A3');
xlswrite(condition, heading_vel, 'min_ang_vel_SD');

xlswrite(condition, ang_acc_7_values, 'ang_acc', 'A2');
xlswrite(condition, heading_acc, 'ang_acc');
xlswrite(condition, ang_acc_sd, 'ang_acc_SD', 'A2');
xlswrite(condition, heading_acc, 'ang_acc_SD');
xlswrite(condition, max_ang_acc_sd, 'max_ang_acc_SD', 'A2');
xlswrite(condition, max_ang_acc_sd_dc, 'max_ang_acc_SD', 'A3');
xlswrite(condition, heading_acc, 'max_ang_acc_SD');
xlswrite(condition, min_ang_acc_sd, 'min_ang_acc_SD', 'A2');
xlswrite(condition, min_ang_acc_sd_dc, 'min_ang_acc_SD', 'A3');
xlswrite(condition, heading_acc, 'min_ang_acc_SD');

xlswrite(condition, hori_vel, 'hori_vel', 'A2');
xlswrite(condition, heading_3, 'hori_vel');


xlswrite(condition, vert_vel, 'vert_vel', 'A2');
xlswrite(condition, heading_3, 'vert_vel');
```

Appendix VIII

MatLab Code for EMG data processing

```
clear
addpath('D:\manchung\fung\kinematic_raw');
fname=input('input the file name ','s');
dd=xlsread(fname);
[dr dc]=size(dd);
x=1:dr;
xx=1:100;
addpath('D:\manchung\fung\emg');
fname=input('input the file name ','s');
fname=strcat(fname,'.xls');
importfile(fname);
ee=data(1:100,:)*1000;
pp=zeros(100,1);
d100=zeros(100,7);
for i= 1:7
pp(1)=1;
for k= 2 :100
pp(k)= pp(k-1)+(dr-1)/99; end
d100(:,i)=spline(x,dd(:,i),pp);  end
%doff=d100(1,:);
%for i =1:7
%    for k=1:100
%    d100(k,i)=d100(k,i)-doff(i);
%    end
%end
dd1=zeros(length(xx),2);
for n=1:dc
for k= 3 : length(xx)-2
dd1(k,n)=(-d100(k-2,n)+8*d100(k-1,n)-8*d100(k+1,n)+d100(k+2,n))/(12/1000); end
end
dd2=zeros(length(xx),2);
for n=1:dc
for k= 3 : length(xx)-2
dd2(k,n)=(-dd1(k-2,n)+8*dd1(k-1,n)-8*dd1(k+1,n)+dd1(k+2,n))/(12/1000); end
end
```

```
%angle vs angle
lm=zeros(7,7);
rm=zeros(7,7);
for i = 1:7
for j= 1:7
cc1=d100(:,i);
cc2=d100(:,j);
[c,lags]=xcorr(cc1,cc2,'coeff');
[r,l]=max(c);
k=lags(l)
lm(i,j)=k;
rm(i,j)=r; end
end
%angle dispment vs emg
le=zeros(7,8);
re=zeros(7,8);
for i = 1:7
for j= 1:8
cc1=d100(:,i);
cc2=ee(:,j);
[c,lags]=xcorr(cc1,cc2,'coeff');
[r,l]=max(c);
k=lags(l)
le(i,j)=k;
re(i,j)=r; end
end
% velocity vs emg
le1=zeros(7,8);
re1=zeros(7,8);
for i = 1:7
for j= 1:8
cc1=dd1(:,i);
cc2=ee(:,j);
[c,lags]=xcorr(cc1,cc2,'coeff');
[r,l]=max(c);
k=lags(l)
le1(i,j)=k;
re1(i,j)=r; end
```

```
end
% velocity vs emg
le2=zeros(7,8);
re2=zeros(7,8);
for i = 1:7
for j= 1:8
cc1=dd2(:,i);
cc2=ee(:,j);
[c,lags]=xcorr(cc1,cc2,'coeff');
[r,l]=max(c);
k=lags(l)
le2(i,j)=k;
re2(i,j)=r; end
end
%plot(xx,d100(:,1),xx,ee(:,1))
```

Appendix IX

The Hong Kong Polytechnic University
Department of Rehabilitation Sciences

Research Project Informed Consent Form

Project title: Validation of three-dimensional (3D) kinematic analysis system

Investigators: Man CHUNG; PhD student, Department of Rehabilitation Science, Hong Kong Polytechnic University

Supervisors: Dr. Simon YEUNG SS; PhD; Associate Professor, Department of Rehabilitation Science, Hong Kong Polytechnic University

Professor Raymond LEE; PhD; Professor of Biomechanics and Head of Sport Science, Roehampton University, London

Details of study:

The aim of this study is to evaluate the validity of the Vicon motion analysis system**.** The results acquired would justify the use of Vicon motion system for 3D kinematic analysis of the fencing actions that are going to be used in latter part of this research project.

Participants who join this study are on voluntary basis and there will be no personal benefit of any form by joining this study. Participants are required to attend one testing session and the procedures will last about 2 hours.

In this study, participants are requested to perform some of the assigned upper limb motions in their best available speed. Two motion analysis systems namely Vicon motion analysis system and X'sens system will be attached to specific body landmarks of the participants for upper limb joint angle motion analysis. The real time angle measured by the two motion analysis systems will then be further computed for validity analysis.

Dangers and Right:

There is no known risk involved except for possible short-term muscle soreness over the assessed arm or low risk of skin allergy due to the reflective markers.

Consent:

I, ___________________________, have been explained the details of this study. I voluntarily consent to participate in this study. I understand that I can withdraw from this study at any time without giving reasons, and my withdrawal will not lead to any punishment or prejudice against me. I am aware of any potential risk in joining this study. I also understand that my personal information will not be disclosed to people who are not related to this study and my name or photograph will not appear on any publications resulted from this study.

I can contact the chief investigator, Mr. Man CHUNG at telephone 9659 for any questions about this study. If I have complaints related to the investigator(s), I can contact Ms Michelle Leung, secretary of Departmental Research Committee, at 2766 5397. I know I will be given a signed copy of this consent form.

Signature (subject): Date:

Signature (witness): Date:

Appendix X

香港理工大學康復治療科學系

參與
三維動態分析系統真確性測試
研究同意書

本人___________________自願及義務參與由香港理工大學博士生鍾惠文先生、香港理工大學康復治療科學系副教授楊世模博士與及英國倫敦 Roehampton University 李潤華教授負責之上述研究。

此項研究的目的是評估 Vicon 動態分析系統的真確性，所得的研究結果將可證明往後以 Vicon 進行劍擊動作分析是可靠的。於實驗進行中，參加者需要以最快的速度進行指定的上肢動作，實驗紀錄員將會以兩個動態分析系統，分別為「Vicon」及「X’ sens」，附繫予參加者的特定身體部位上，以收集上肢動態角度的數據。參與者只需進行一次測試，而每次測試大約為期兩小時。

根據研究人員的專業知識，這些測試除了可能導致肌肉疲勞外，不會為你帶來任何不良的後果。而將貼在參加者身上的光標亦不會引起嚴重的皮膚刺激或過敏反應。此外，參加者亦需明白其參與純屬義務性質，參與者將不會得到任何形式的利益，故即使參與者欲拒絕繼續參與有關研究，亦不需付上任何責任或導致任何利益上之損失。

本人在此項研究過程中，可提出任何有關研究程序的疑問，並且已應該得到上述之調查負責人員的回應和解釋。假使上述調查之負責人員未能對本人之提問給予滿意的答覆，本人可就有關查詢致電 9659 聯絡鍾惠文先生。若本人對這項研究有任何不滿，亦可以致電 2766 5397 聯絡康復治療科學系研究委員會秘書梁小姐。同時，本人亦知悉此研究的結果，除可能作綜合報告外，本人之個人資料將會保密。本人已經閱畢及完全明白此同意書之內容，並已收到此同意書的副本乙份以作參考。

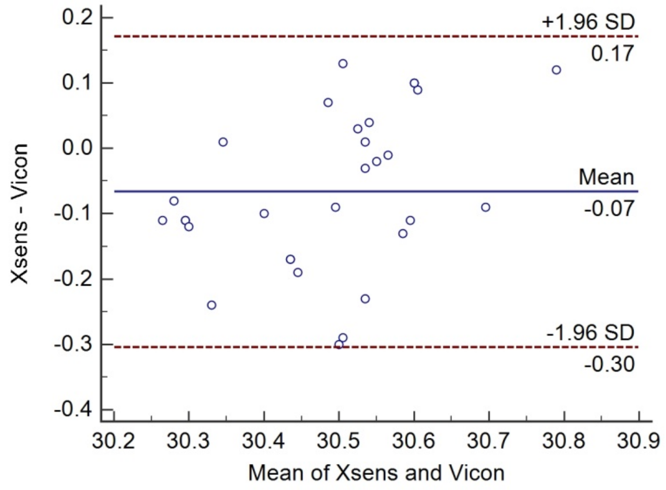
______________________ ______________________
參加者 見證人

______________________ ______________________
日期 日期

Appendix XI

Bland and Altman plots for Xsens/ Vicon system measurement

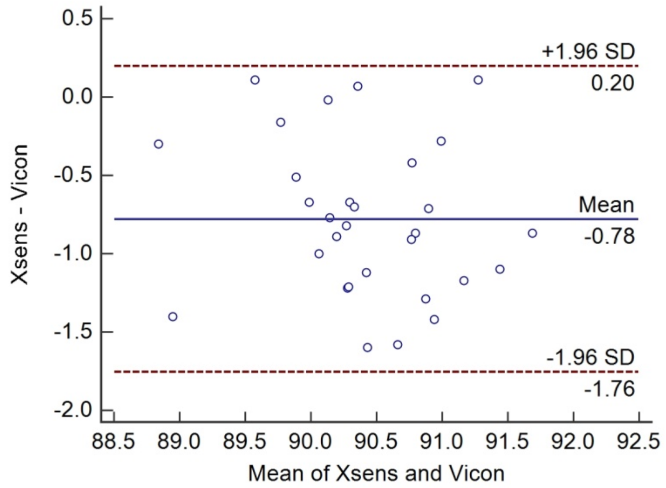
a) Shoulder flexion



Shoulder flexion at 30°

Mean difference: 0.07

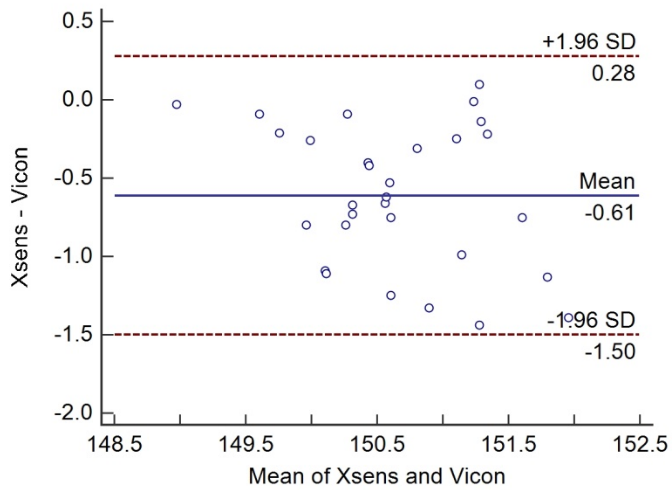
95% CI: -0.11 to -0.02



Shoulder flexion at 90°

Mean difference: -0.78

95% CI: -0.97 to -0.59

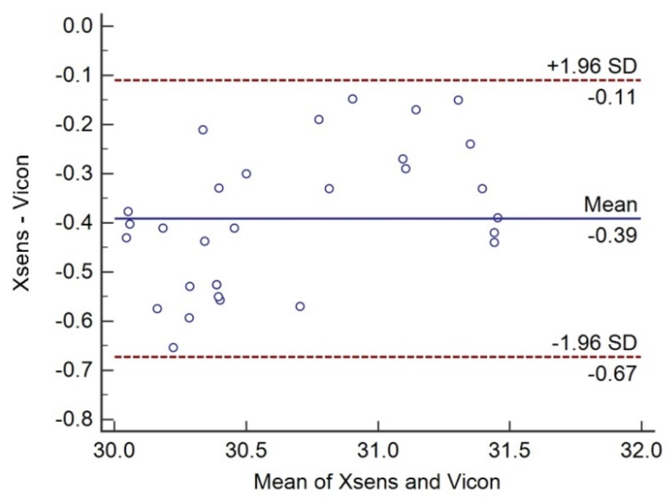
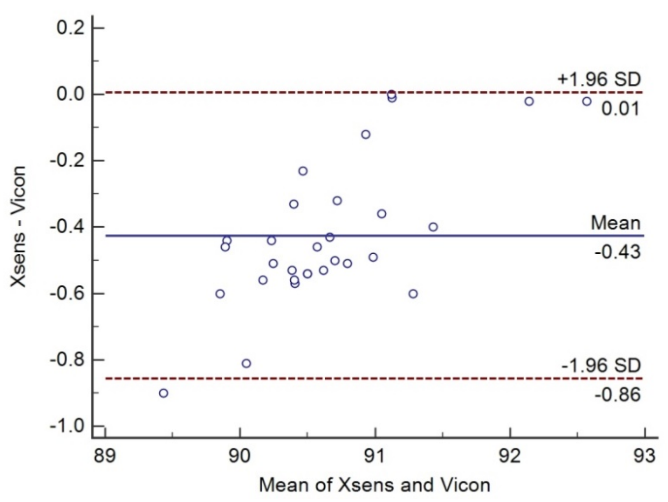
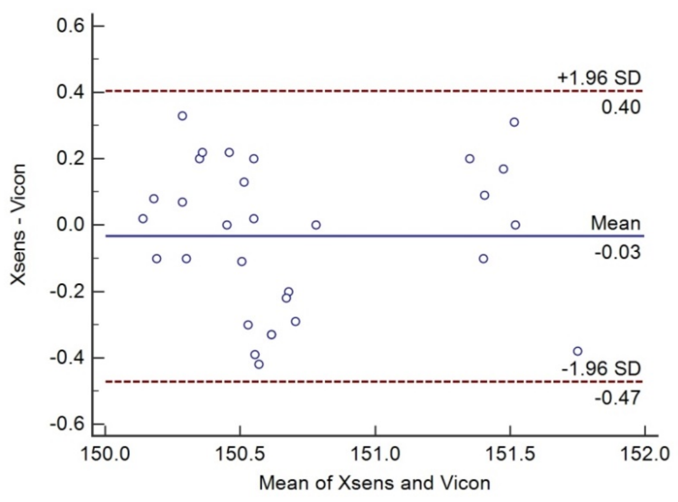


Shoulder flexion at 150°

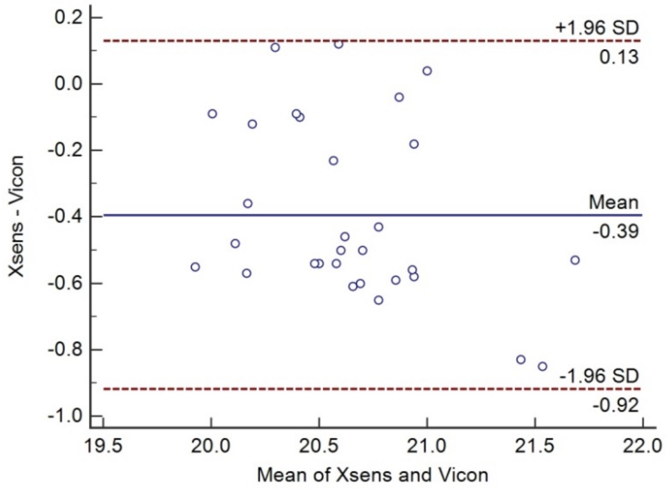
Mean difference: -0.61

95% CI: -0.78 to -0.44

b) Shoulder abduction

	**Shoulder abduction at 30°** Mean difference: -0.39 95% CI: -0.45 to -0.34
	**Shoulder abduction at 90°** Mean difference: -0.43 95% CI: -0.51 to -0.34
	**Shoulder abduction at 150°** Mean difference: -0.03 95% CI: -0.12 to 0.05

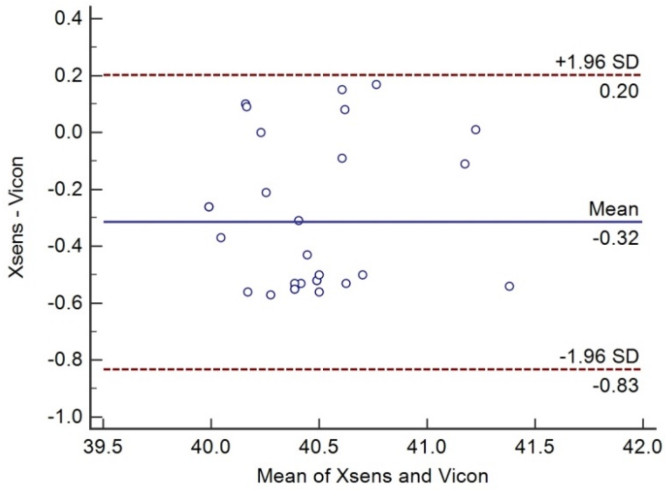
c) Shoulder rotation



Shoulder external rotation at 20°

Mean difference: -0.40

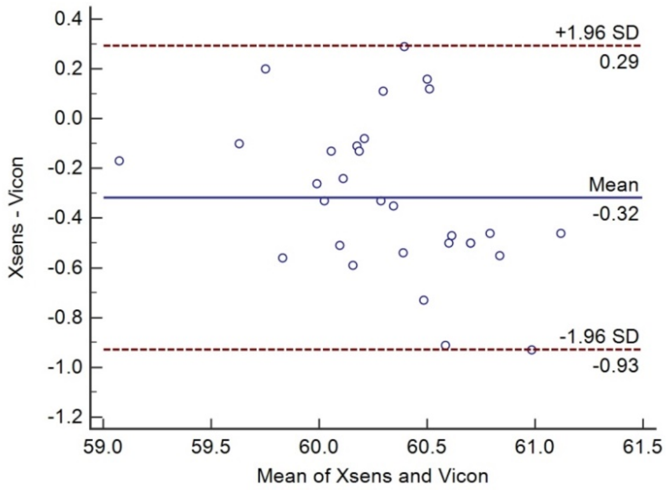
95% CI: -0.30 to -0.50



Shoulder external rotation at 40°

Mean difference: -0.32

95% CI: -0.41 to -0.22

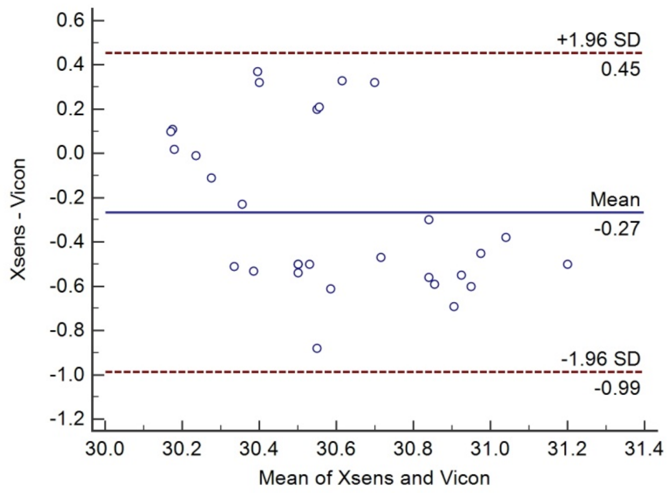


Shoulder external rotation at 60°

Mean difference: -0.32

95% CI: -0.44 to -0.20

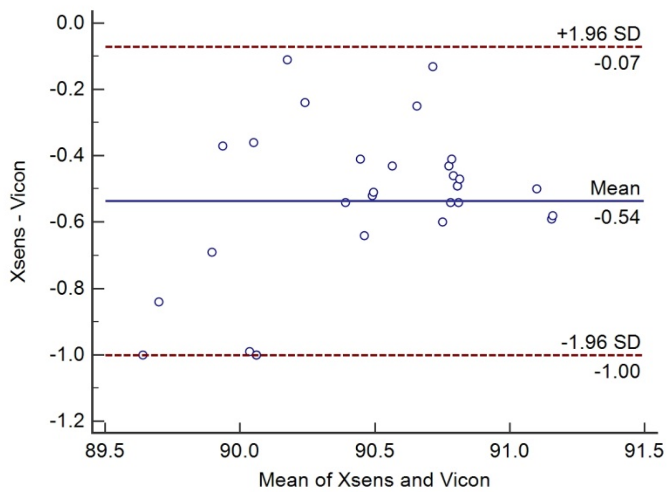
d) Elbow flexion



Elbow flexion at 30°

Mean difference: -0.27

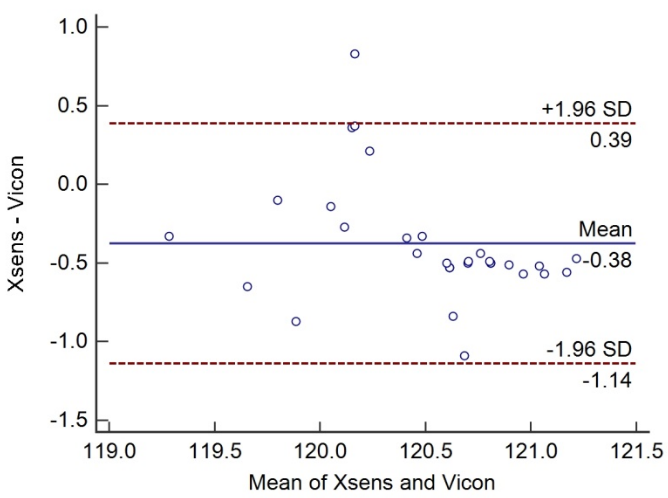
95% CI: -0.40 to -0.13



Elbow flexion at 90°

Mean difference: -0.54

95% CI: -0.63 to -0.45

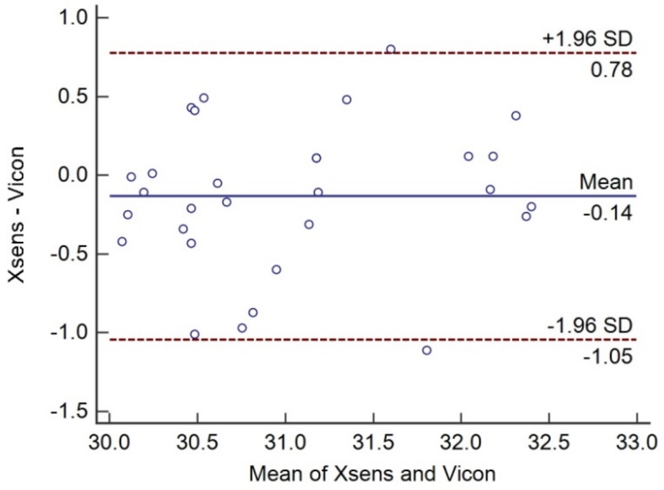


Elbow flexion at 120°

Mean difference: -0.38

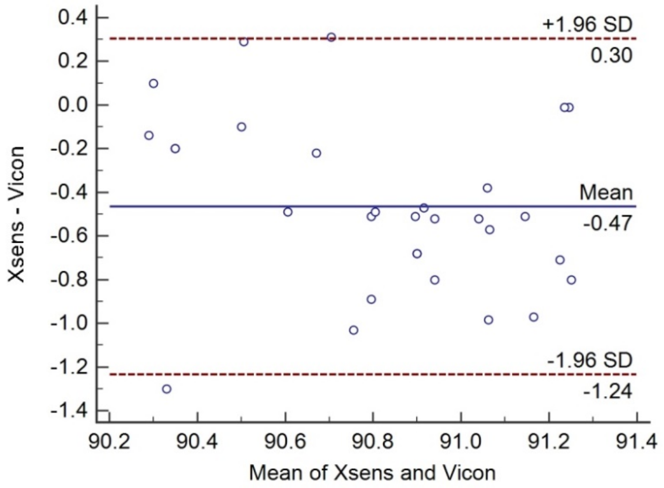
95% CI: -0.52 to -0.23

e) Forearm supination



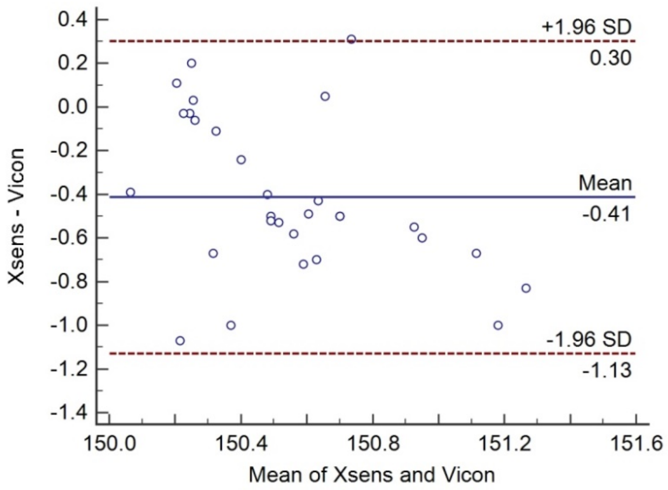
Forearm supination at 30°

Mean difference: -0.14
95% CI: -0.31 to 0.04



Forearm supination at 90°

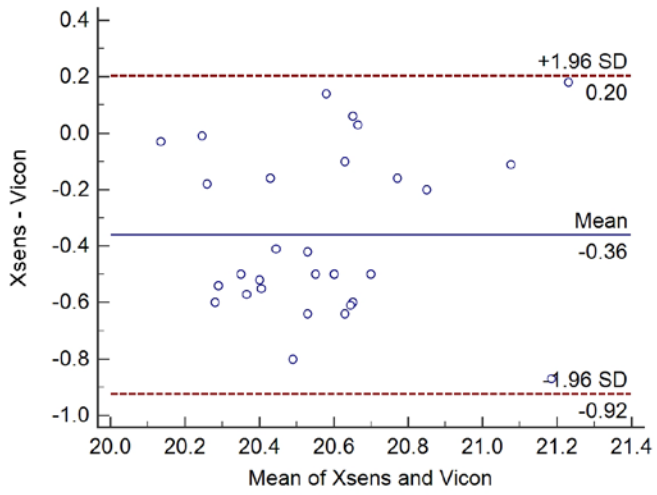
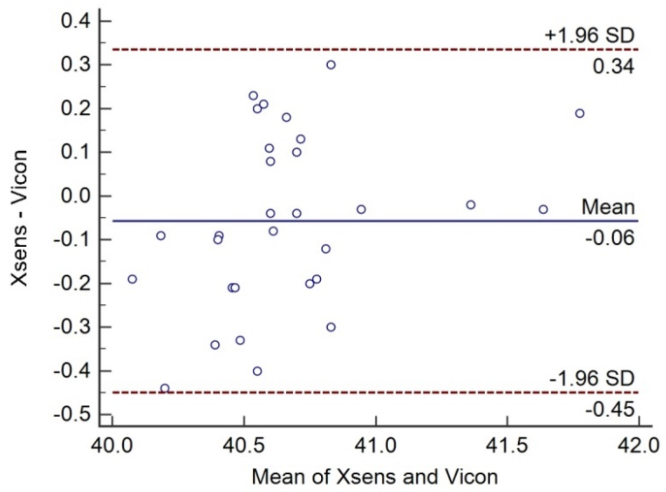
Mean difference: -0.47
95% CI: -0.61 to -0.32



Forearm supination at 150°

Mean difference: -0.41
95% CI: -0.55 to -0.28

f) Wrist flexion

	**Wrist flexion at 20°** Mean difference: -0.36 95% CI: -0.47 to -0.25
	**Wrist flexion at 40°** Mean difference: -0.06 95% CI: -0.13 to 0.02

Appendix XII

The Hong Kong Polytechnic University
Department of Rehabilitation Sciences

Research Project Informed Consent Form

Project title: The effect of fencing posture and disability levels on upper limb kinematic

Investigators: Man CHUNG; PhD student, Department of Rehabilitation Science, Hong Kong Polytechnic University

Supervisors: Dr. Simon YEUNG SS; PhD; Associate Professor, Department of Rehabilitation Science, Hong Kong Polytechnic University

Professor Raymond LEE; PhD; Professor of Biomechanics and Head of Sport Science, Roehampton University, London

Details of study:

The aim of this study is to evaluate the effect of fencing posture and disability level on the upper limb kinematics of the wheelchair fencers. The findings will provide information for better understanding the movement science of wheelchair fencing and the possible associated injury mechanism of upper limb extremities among wheelchair fencers.

Participants who join this study are on voluntary basis and there will be no personal benefit of any form by joining this study. Participants are required to attend one testing session and the procedures will last about 2 hours.

This project will focus on the fencing motion analysis in sitting position. Skin markers will be attached to the sword and specific landmarks of each participant for measuring the upper limb kinematics (i.e. joint angle, speed and acceleration) during fencing. Subsequent to 5-minutes warm-up, participant will be instructed to perform 5 trails of the 6 assigned fencing techniques with the sword hit accurately on the pre-set target.

Dangers and Right:

There is no known risk involved except for possible short-term muscle soreness over the tested arm or low risk of skin allergy due to the reflective markers.

Consent:

I, ___________________________, have been explained the details of this study. I voluntarily consent to participate in this study. I understand that I can withdraw from this study at any time without giving reasons, and my withdrawal will not lead to any punishment or prejudice against me. I am aware of any potential risk in joining this study. I also understand that my personal information will not be disclosed to people who are not related to this study and my name or photograph will not appear on any publications resulted from this study.

I can contact the chief investigator, Mr. Man CHUNG at telephone 9659 for any questions about this study. If I have complaints related to the investigator(s), I can contact Ms Michelle Leung, secretary of Departmental Research Committee, at 2766 5397. I know I will be given a signed copy of this consent form.

Signature (subject): Date:

Signature (witness): Date:

香港理工大學康復治療科學系

參與
「殘障程度及坐立穩定性」對劍擊上肢動態的影響
研究同意書

本人____________________自願及義務參與由香港理工大學博士生鍾惠文先生、香港理工大學康復治療科學系副教授楊世模博士與及英國倫敦 Roehampton University 李潤華教授負責之上述研究。

此項研究的主要目的是分析不同劍擊動作時的上肢動態（關肢角度，速度與加速度等）。收集數據的方式將會以坐姿進行，參加者的不同身體部位將會貼上反光標記以收集動態數據。經過五分鐘熱身後，參加者需要以劍作出不同的劍擊動作，而每個動作將要重覆五次。實驗進行時，參與者需要以全力作出每一個動作，準確地擊中預設的目標。在一般情況下，實驗將會在兩小時內完成。

透過這個實驗，我們希望能取得重要的數據，了解殘障程度及坐立穩定性對輪椅劍擊運動員上肢動態的影響，從而理解其相關的受傷機制。根據研究人員的專業知識，這些測試除了可能導致肌肉疲勞外，不會為你帶來任何不良的後果。而將貼在參加者身上的光標亦不會引起嚴重的皮膚刺激或過敏反應。

此外，參與者亦需明白其參與純屬義務性質，參與者將不會得到任何形式的利益，故即使參與者欲拒絕繼續參與有關研究，亦不需付上任何責任或導致任何利益上之損失。本人在此項研究過程中，可提出任何有關研究程序的疑問，並且已應該得到上述之調查負責人員的回應和解釋。假使上述調查之負責人員未能對本人之提問給予滿意的答覆，本人可就有關查詢致電 9659 　　　聯絡鍾惠文先生。若本人對這項研究有任何不滿，亦可以致電 2766 5397 聯絡康復治療科學系研究委員會秘書梁小姐。同時，本人亦知悉此研究的結果，除可能作綜合報告外，本人之個人資料將會保密。本人已經閱畢及完全明白此同意書之內容，並已收到此同意書的副本乙份以作參考。

______________________	______________________
參加者	見證人
______________________	______________________
日期	日期

REFERENCE

1. Aggeloussis N, Gourgoulis V, Sertsou M, Giannakou E, Mavromatis G. (2007) Repeatability of electromyographic waveforms during the Naeryo Chagi in taekwondo. *Journal of Sports Science & Medicine*, 6 (CSSI-2), 6-9.

2. Alexander RM. (1991) Optimum timing of muscle activation for simple models of throwing. *Journal of Theoretical Biology,* 150, 349-372.

3. An K-N, Browne AO, Korinek S, Tanaka S, Morrey BF. (1991) Three-dimensional kinematics of glenohumeral elevation. *Journal of Orthopaedic Research*, 9, 143–149.

4. Anglin C, & Wyss U P. (2000) Review of arm motion analyses. *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, 214 (5), 541-555.

5. Aquili A, Tancredi V, Triossi T, De Sanctis D, Padua E, D'Arcangelo G, Melchiorri G. (2013) Performance analysis in saber. *The Journal of Strength & Conditioning Research*, 27 (3), 624-630.

6. Athanasopoulos S, Mandalidis D, Tsakoniti A, Athanasopoulos I, Strimpakos N, Papadopoulos E, Kapreli E. (2009) The 2004 Paralympic Games: Physiotherapy Services in the Paralympic Village Polyclinic. *Open Sports Medicine Journal*, *3*, 1-8.

7. Bachschmidt RA, Harris GF, Simoneau GG. (2001) Walker-assisted gait in rehabilitation: a study of biomechanics and instrumentation. *Neural Systems and Rehabilitation Engineering, IEEE Transactions on*, 9 (1), 96-105.

8. Bahr R, Krosshaug T. (2005) Understanding injury mechanisms: a key component of preventing injuries in sport. *British Journal of Sports Medicine*, 39 (6), 324-329.

9. Bahr R, Lian O, Bahr I. (1997) A twofold reduction in the incidence of acute ankle sprains in volleyball after the introduction of an injury prevention program – a prospective cohort study. *Scandinavian Journal of Medicine & Science in Sport,* 7 (3), 172-177.

10. Basmajian JV, De Luca CJ. (1985) *Muscles alive: their Function revealed by Electromyography*. Baltimore: Williams & Wilkins.

11. Bedi G. (2011) Shoulder injury in athletes. *Journal of Clinical Orthopaedics and Trauma*, 2 (2), 85-92.

12. Bennell KL, Crossley K. (1996) Musculoskeletal injuries in track and field: incidence, distribution and risk factors. *Australian Journal of Science and Medicine in Sport,* 28 (3), 69–75.

13. Beres-Jones JA, Harkema SJ. (2004) The human spinal cord interprets velocity-dependent afferent input during stepping. *Brain,* 127 (10), 2232-2246.

14. Bergmann JH, Mayagoitia RE, Smith I. (2009) A portable system for collecting anatomical joint angles during stair ascent: a comparison with an optical tracking device. *Dynamic Medicine*, 8, 1-7.

15. Bernstein N. (1967) *The coordination and regulation of movements*. Oxford: Pergamon Press.

16. Biryukova E V, Roby-Brami A, Frolov A A, Mokhtari M. (2000) Kinematics of human arm reconstructed from spatial tracking system recordings. *Journal of Biomechanics*, 33 (8), 985-995.

17. Bland JM, Altman DG. (1986) Statistical methods for assessing agreement between two methods of clinical measurement. *The Lancet*, 327 (8476), 307-310.

18. Bland JM, Altman DG. (2010) Statistical methods for assessing agreement between two methods of clinical measurement. *International Journal of Nursing Studies*, 47, 931-936.

19. Blauwet C, Willick SE. (2012) The Paralympic Movement: using sports to promote health, disability rights, and social integration for athletes with disabilities. *PM&R*, 4 (11), 851-856.

20. Bolgla LA, Uhl TL. (2007) Reliability of electromyographic normalization methods for evaluating the hip musculature. *Journal of Electromyography and*

Kinesiology, 17 (1), 102-111.

21. Boninger ML, Koontz AM, Sisto SA, Dyson-Hudson TA, Chang M, Price R, Cooper RA. (2005) Pushrim biomechanics and injury prevention in spinal cord injury: Recommendations based on CULP-SCI investigation. *Journal of Rehabilitation Research and Development,* 42 (3), 9-20.

22. Bouisset S, Do MC. (2008) Posture, dynamic stability, and voluntary movement. *Neurophysiologie Clinique/Clinical Neurophysiology*, 38 (6), 345-362.

23. Bouisset S, Zattara M. (1981) A sequence of postural movements precedes voluntary movement. *Neuroscience Letters*, 22 (3), 263-270.

24. Bronner S, Agraharasamakulam S, Ojofeitimi S. (2010) Reliability and validity of electrogoniometry measurement of lower extremity movement. *Journal of Medical Engineering & Technology*, 34 (3), 232-242.

25. Burnham R, Newell E, Steadward R. (1991) Sports medicine for the physically disabled: the Canadian team experience at the 1988 Seoul Paralympic games. *Clinical Journal of Sports Medicine,* 1, 193-196.

26. Chen CL, Yeung KT, Bih LI, Wang CH, Chen MI, Chien JC. (2003) The relationship between sitting stability and functional performance in patients with paraplegia. *Archives of Physical Medicine and Rehabilitation*, 84 (9), 1276-1281.

27. Chiari L, Croce UD, Leardini A, Cappozzo A. (2005) Human movement analysis using stereophotogrammetry: Part 2: Instrumental errors. *Gait & posture*, 21 (2), 197-211.

28. Cholewicki J, Silfies SP, Shah RA, Greene HS, Reeves NP, Alvi K, Goldberg B. (2005) Delayed trunk muscle reflex responses increase the risk of low back injuries. *Spine,* 30 (23), 2614-2620.

29. Chow JW, Mindock LA. (1999) Discus throwing performances and medical classification of wheelchair athletes. *Medicine and Science in Sports and Exercise*, 31 (9), 1272-1279.

30. Chow J, Kuenster AF, Lim YT. (2003) Kinematic analysis of javelin throw

performed by wheelchair athletes on different functional classes. *Journal of Sports Science and Medicine*, 2, 36-46.

31. Chung WM. (2008) Classification for wheelchair fencing. *Sports Physio,* 2, 23-25.

32. Clarke GM, Cooke D. (2005) *A Basic Course in Statistics. 5th ed*. London, United Kingdom: Oxford University Press.

33. Comerford MJ, Mottram SL. (2001) Movement and stability dysfunction–contemporary developments. *Manual Therapy,* 6 (1), 15-26.

34. Cools AM, Witvrouw EE, Declercq GA, Danneels LA, Cambier DC. (2003) Scapular muscle recruitment patterns: trapezius muscle latency with and without impingement symptoms. *American Journal of Sports Medicine*, 31 (4), 542-549.

35. Cooper RA, Quatrano LA, Axelson PW, Harlan W, Stineman M, Franklin B, Krause JS, Bach J, Chambers H, Chao EY, Alexander M, Painter P. (1999) Research on physical activity and health among people with disabilities: a consensus statement. *Journal of Rehabilitation Research and Development*, 36 (2), 142-154.

36. Cronin J, McNAIR P, Marshall R. (2003) Lunge performance and its determinants. *Journal of Sports Sciences*, 21 (1), 49-57.

37. Cuesta-Vargas AI, Galán-Mercant A, Williams JM. (2010) The use of inertial sensors system for human motion analysis. *Physical Therapy Reviews*, 15 (6), 462-473.

38. Curtis KA, Black K. (1999) Shoulder pain in female wheelchair basketball players. *Journal of Orthopaedic & Sports Physical Therapy*, 29 (4), 225-231.

39. Curtis KA, Dillon DA. (1985) Survey of wheelchair athletic injuries: common patterns and prevention. *Spinal Cord*, 23 (3), 170-175.

40. Curtis KA, McClanahan S, Hall KM, Dillon D, Brown KF. (1986) Health, vocational, and functional status in spinal cord injured athletes and nonathletes. *Archives of Physical Medicine and Rehabilitation*, 67, 862–865

41. Dankaerts W, O'Sullivan PB, Burnett AF, Straker LM, Danneels LA. (2004) Reliability of EMG measurements for trunk muscles during maximal and sub-maximal voluntary isometric contractions in healthy controls and CLBP patients. *Journal of Electromyography and Kinesiology*, 14 (3), 333-342.

42. Davis JL, Growney ES, Johnson ME, Iuliano BA, An KN. (1998) Three-dimensional kinematics of shoulder complex during wheelchair propulsion. *Journal of Rehabilitation Research and Development,* 35 (1), 61-72.

43. De Luca CJ. (1997) The use of surface electromyography in biomechanics. *Journal of Applied Biomechanics*, 13, 135-163.

44. Dean C, Shepherd R, Adams R. (1999) Sitting balance I: trunk–arm coordination and the contribution of the lower limbs during self-paced reaching in sitting. *Gait & Posture*, 10 (2), 135-146.

45. Della Croce U, Leardini A, Chiari L, Cappozzo A. (2005) Human movement analysis using stereophotogrammetry: Part 4: assessment of anatomical landmark misplacement and its effects on joint kinematics. *Gait & posture*, 21 (2), 226-237.

46. DePauw KP. (2012) A historical perspective of the Paralympic Games. *Journal of Physical Education, Recreation & Dance*, 83 (3), 21-31.

47. DePauw KP, Gavron SJ. (2005) *Disability Sport*. 2nd ed. Champaign, IL: Human Kinetics.

48. Desroches G, Gagnon D, Nadeau S, Popovic MR. (2013) Effects of sensorimotor trunk impairments on trunk and upper limb joint kinematics and kinetics during sitting pivot transfers in individuals with a spinal cord injury. *Clinical Biomechanics*, 28 (1), 1-9.

49. Do MC, Bouisset S, Moynot C. (1985) Are paraplegics handicapped in the execution of a manual task? *Ergonomics*, 28 (9), 1363-1375.

50. Doorenbosch CA, Harlaar J, Veeger DH. (2003) The globe system: an unambiguous description of shoulder positions in daily life movement. *Journal of Rehabilitation Research and Development*, 40 (1), 47-55.

51. Dubowsky SR, Sisto SA, Langrana NA. (2009) Comparison of kinematics, kinetics, and EMG throughout wheelchair propulsion in able-bodied and persons with paraplegia: an integrative approach. *Journal of Biomechanical Engineering*, 131 (2), 1-8.

52. Engebretsen L, Soligard T, Steffen K, Alonso JM, Aubry M, Budgett R, Renström PA. (2013) Sports injuries and illnesses during the London Summer Olympic Games 2012. *British Journal of Sports Medicine*, 47 (7), 407-414.

53. Fagarasanu M, Kumar S. (2002) Measurement instruments and data collection: a consideration of constructs and biases in ergonomics research. *International Journal of Industrial Ergonomics*, 30 (6), 355-369.

54. Fagher, K., & Lexell, J. (2014). Sports-related injuries in athletes with disabilities. *Scandinavian Journal of Medicine & Science in Sports*. DOI: 10.1111/sms.12175

55. Farina D, Merletti R, Enoka RM. (2004) The extraction of neural strategies from surface EMG. *Journal of Applied Physiology,* 96, 1486-1495.

56. Ferrara MS, Buckley WE, McCann BC, Limbird TJ, Powell JW, Robl R. (1992). The injury experience of the competitive athlete with a disability: prevention implications. *Medicine and Science in Sports and Exercise*, 24 (2), 184-188.

57. Ferrara MS, Buckley WE. (1996) Athletes with disabilities injury registry. *Adapted Physical Activity Quarterly*, 13, 50-60.

58. Ferrara MS, Davis RW. (1990) Injuries to elite wheelchair athletes. *Spinal Cord*, 28 (5), 335-341.

59. Ferrara MS, Petterson CL. (2000) Injuries to athletes with disabilities: identifying injury pattern. *Sports Medicine*, 30 (2), 137-143.

60. Fleisig GS, Barrentine SW, Escamilla RF, Andrews JR. (1996) Biomechanics of overhand throwing with implications for injuries. *Sports Medicine*, 21 (6), 421-437.

61. Fleisig GS, Barrentine SW, Zheng N, Escamilla RF, Andrews JR. (1999) Kinematic and kinetic comparison of baseball pitching among various levels of

development. *Journal of Biomechanics*, 32 (12), 1371-1375.

62. Forner-Cordero A, Mateu-Arce M, Forner-Cordero I. Alcántara,E, Moreno JC, Pons JL. (2008) Study of the motion artefacts of skin-mounted inertial sensors under different attachment conditions. *Physiological Measurement*, 29, 21-31.

63. Forouzan AS, Mahmoodi A, Shushtari ZJ, Salimi Y., Sajjadi H, Mahmoodi Z. (2013) Perceived Social Support Among People With Physical Disability. *Iranian Red Crescent Medical Journal*, 15 (8), 663.

64. Frère J, Göpfert B, Nüesch C, Huber C, Fischer M, Wirz D, Friederich NF. (2011) Kinematical and EMG-classifications of a fencing attack. *International Journal of Sports Medicine*, 32 (1), 28-34.

65. Frère J, Nüesch C, Fischer M, Göpfert B, Wirz D, Friederich NF. (2008) Shoulder muscles coordination of the weapon side during a fencing attack: the Fleche. *Journal of Biomechanics*, 41, S58.

66. Friedli WG, Hallett M, Simon SR. (1984) Postural adjustments associated with rapid voluntary arm movements 1. Electromyographic data. *Journal of Neurology, Neurosurgery & Psychiatry*, 47 (6), 611-622.

67. Frigo C, Shiav R. (2004) Applications in movement and gait analysis. *Electromyography: Physiology, Engineering, and Noninvasive Applications*, 381-401.

68. Fung YK, Chan DKC, Caudwell KM, Chow BC. (2013) Is the wheelchair fencing classification fair enough? A kinematic analysis among world-class wheelchair fencers. *European Journal of Adapted Physical Activity*, 6 (1), 17-29.

69. Gagnon D, Nadeau S, Gravel D, Noreau L, Larivière C, Gagnon D. (2003) Biomechanical analysis of a p osterior transfer maneuver on a l evel surface in individuals with high and low-level spinal cord injuries. *Clinical biomechanics*, 18 (4), 319-331.

70. Gagnon D, Nadeau S, Noreau L, Dehail P, Piotte F. (2008) Comparison of peak shoulder and elbow mechanical loads during weight-relief lifts and sitting pivot transfers among manual wheelchair users with spinal cord injury. *Journal of*

Rehabilitation Research and Development, 45 (6), 863-873.

71. Gatts GC, Camp EM. (1979) Physiological characteristics of trained Australian paraplegic and tetraplegic subjects. *Medicine and Science in Sports and Exercise*, 11, 256–259.

72. Gauthier C, Gagnon D, Grangeon M, Jacquemin G, Nadeau S, Masani K, Popovic MR. (2013) Comparison of multidirectional seated postural stability between individuals with spinal cord injury and able-bodied individuals. *Journal of Rehabilitation Medicine*, 45 (1), 47-54.

73. Gebhardt DL. (1981) Force and power production in the fencing lunge. *Journal of Biomechanics*, 14 (7), 494.

74. Gholipour M, Tabrizi A, Farahmand F. (2008) Kinematics analysis of lunge fencing using stereophotogrametry. *World Journal of Sport Sciences*, 1 (1), 32-37.

75. Giroux B, Lamontagne M. (1990) Comparisons between surface electrodes and intramuscular wire electrodes in isometric and dynamic conditions. *Electromyography and Clinical Neurophysiology*, 30 (7), 397-405.

76. Goodwin PC, Koorts K, Mark R, Mai S, Morrissey MC, Hooper DM. (1999) Reliability of leg muscle electromyography in vertical jumping. *European Journal of Applied Physiology,* 79, 374-378.

77. Grea H, Desmurget M, Prablanc C. (2000) Postural invariance in three-dimensional reaching and grasping movements. *Experimental Brain Research*, 134, 155-162.

78. Groah SL, Lanig IS. (2000) Neuromusculoskeletal syndromes in wheelchair athletes. *Seminar in Neurology,* 20, 201-208.

79. Grood ES, Suntay WJ. (1983) A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *Journal of Biomechanical Engineering*, 105, 136-144.

80. Groppal JL. (1992) *High tech tennis*, 2nd edition, Champaign, IL, Human Kinetics Publishers.

81. Guilhem G, Giroux C, Couturier A, Chollet D, Rabita G. (2014) Mechanical and Muscular Coordination Patterns during a High-Level Fencing Assault. *Medicine and Science in Sports and Exercise*, 46 (2), 341-350.

82. Gutiérrez-Dávila M, Rojas FJ, Antonio R, Navarro E. (2013) Effect of uncertainty on the reaction response in fencing. *Research Quarterly for Exercise and Sport*, 84 (1), 16-23.

83. Harmenberg J, Ceci R, Barvestad P, Hjerpe K, Nyström J. (1991) Comparison of different tests of fencing performance. *International Journal of Sports Medicine*, 12 (06), 573-576.

84. Harmer PA. (1996) Disability Sports. *Epidemiology of Sports Injuries*, 12, 161-175.

85. Harmer PA. (2008) Incidence and characteristics of time-loss injuries in competitive fencing: a prospective, 5-year study of national competitions. *Clinical Journal of Sport Medicine*, 18 (2), 137-142.

86. Harris GF, Smith PA. (1996) *Human Motion Analysis*. New York: IEEE Press.

87. Hassan SEA, Klauck J. (1998) Kinematics of lower and upper extremity motions during the fencing lunge: results and training implications. Abstracts from ISBS symposium. www.isbs 98.uni-konstanz.de/abstracts/klauckl.pdf

88. Hingtgen B, McGuire J R, Wang M, Harris GF. (2006) An upper extremity kinematic model for evaluation of hemiparetic stroke. *Journal of Biomechanics*, 39 (4), 681-688.

89. Hirashima M, Kadota H, Sakurai S, Kudo K, Ohtsuki T. (2002) Sequential muscle activity and its functional role in the upper extremity and trunk during overarm throwing. *Journal of Sports Science,* 20 (4), 301-310.

90. Hodges PW, Richardson CA. (1997) Feedforward contraction of transversus abdominis is not influenced by the direction of arm movement. *Experimental Brain Research*, 114 (2), 362-370.

91. Hodges PW, Richardson CA. (1999) Altered trunk muscle recruitment in people

with low back pain with upper limb movement at different speeds. *Archives of Physical Medicine and Rehabilitation*, 80 (9), 1005-1012.

92. Hodges PW, Moseley GL, Gabrielsson A, Gandevia SC. (2003) Experimental muscle pain changes feedforward postural responses of the trunk muscles. *Experimental Brain Research*, 151 (2), 262-271.

93. Hsu WL, Yang YR, Hong CT, Wang RY. (2005) Ankle muscle activation during functional reach in hemiparetic and healthy subjects. *American Journal of Physical Medicine & Rehabilitation*, 84 (10), 749-755.

94. Hug François. (2011) "Can muscle coordination be precisely studied by surface electromyography?" *Journal of Electromyography and Kinesiology,* 21 (1), 1-12.

95. Hughes PC. (2012) Measurement of subacromial impingement of the rotator cuff. *Journal of Science and Medicine in Sport*, 15, 2-7.

96. International Paralympic Committee – Paralympic Games Summer. Retrieved 30 April 2014, from http://www.paralympic.org/paralympic-games/summer

97. International Perspectives on S pinal Cord Injury. Geneva, World Health Organization & The international Spinal Cord Society, 2013.

98. International Wheelchair & Amputee Sports Federation. Wheelchair fencing - History of wheelchair fencing. Retrieved May 1, 2014, from http://www.iwasf.com/iwasf/index.cfm/sports/iwas-wheelchair-fencing/history/

99. International Wheelchair & Amputee Sports Federation. (2001) International wheelchair fencing rules for competition: Book 2 – Classification rules. IWAS, Bucks, UK.

100. Janda DH. (2003) The prevention of baseball and softball injuries. *Clinical Orthopedics and Related Research,* 409, 20-28.

101. Jasiewicz JM, Allum JH, Middleton JW, Barriskill A, Condie P, Purcell B. (2006) Gait event detection using linear accelerometers or angular velocity transducers in able-bodied and spinal-cord injured individuals. *Gait & Posture,* 24 (4), 502-509.

102. Jaspers E, Feys H, Bruyninckx H, Harlaar J, Molenaers G, Desloovere K. (2011) Upper limb kinematics: development and reliability of a clinical protocol for children. *Gait & Posture*, 33 (2), 279-285.

103. Kadaba MP, Ramakrishnan HK, Wootten ME. (1990) Measurement of lower extremity kinematics during level walking. *Journal of Orthopaedic Research*, 8 (3), 383-392.

104. Kadaba MP, Ramakrishnan HK, Wootten ME, Gainey J, Gorton G, Cochran GVB. (1989) Repeatability of kinematic, kinetic and electromyographic data in normal adult gait. *Journal of Orthopaedic Research,* 7, 849-860.

105. Kaminski TR. (2007) The coupling between upper and lower extremity synergies during whole body reaching. *Gait & Posture*, 26 (2), 256-262.

106. Kaminski TR, Bock C, Gentile AM. (1995) The coordination between trunk and arm motion during pointing movements. *Experimental Brain Research,* 106 (3), 457-466.

107. Karamanidis K, Arampatzis A, Bruggemann GP. (2004) Reproducibility of electromyography and ground reaction force during various running techniques. *Gait & Posture,* 19, 115-123.

108. Kegel B, Malchow D. (1994) Incidence of injury in amputees playing soccer. *Palaestra*, 10 (2), 50-54.

109. Kelly BT, Backus SI, Warren RF, Williams RJ. (2002) Electromyographic analysis and phase definition of the overhead football throw. *The America Journal of Sports Medicine,* 30 (6), 837-844.

110. Kendall FP, McCreary EK, Provance PG. (1993) *Muscles, Testing and Function: With Posture and Pain*, Baltimore.

111. Kibler WB. (1994) Clinical biomechanics of the elbow in tennis: implications for evaluation and diagnosis. *Medicine and Science in Sports and Exercise*, 26 (10), 1203-1206.

112. Kibler WB. (1998) The role of the scapula in athletic shoulder function. *The American Journal of Sports Medicine*, 26 (2), 325-337.

113. Klenck C, Gebke K. (2007) Practical management: common medical problems in disabled athletes. *Clinical Journal of Sport Medicine*, 17(1), 55-60.

114. Kokito AS, Jabe FW, Pink MM, J Perry, J Brault, (1998) Electromyographic analysis of shoulder function during the volleyball serve and spike. *Journal of Shoulder and Elbow Surgery, 7(3), 256-263.*

115. Kollmitzer J, Ebenbichler G, Kopf A. (1999) Reliability of surface electromyographic measurements. *Journal of Electromyography and Kinesiology,* 110 (4), 725-34.

116. Koontz AM, Cooper RA, Boninger ML, Souza AL, Fay BT. (2002) Shoulder kinematics and kinetics during two speeds of wheelchair propulsion. *Journal of Rehabilitation Research and Development,* 39 (6), 635-649.

117. Kronberg M, Nemeth G, Brostrom LA. (1990) Muscle activity and coordination in the normal shoulder: an electromyographic study. *Clinical Orthopaedics and Related Research*, 257, 76-85.

118. Lam J, Chien P, Au YY. (1995) Shoulder problems in elite disabled sportsmen. *Unpublished data*.

119. Leardini A, Chiari L, Croce UD, Cappozzo A. (2005) Human movement analysis using stereophotogrammetry: Part 3. Soft tissue artifact assessment and compensation. *Gait & posture*, 21 (2), 212-225.

120. Lee RYW. (2001) Kinematics of rotational mobilization of the lumbar spine. *Clinical Biomechanics,*16, 481-488.

121. Lee RYW, Laprade J, Fung EH. (2003) A real-time gyroscopic system for three-dimensional measurement of lumbar spine motion. *Medical Engineering & Physics*, 25 (10), 817-824.

122. Lee RYW, Wong TKT. (2002) Relationship between the movements of the lumbar spine and hip. *Human Movement Science*, 21, 481-494.

123. Lempereur M, Brochard S, Mao L, Rémy-Néris O. (2012) Validity and reliability of shoulder kinematics in typically developing children and children with hemiplegic cerebral palsy. *Journal of Biomechanics*, 45 (11), 2028-2034.

124. Levin M F, Michaelsen SM, Cirstea C M, Roby-Brami A. (2002). Use of the trunk for reaching targets placed within and beyond the reach in adult hemiparesis. *Experimental Brain Research*, 143 (2), 171-180.

125. Li L, Caldwell, GE. (1999) Coefficient of cross correlation and the time domain correspondence. *Journal of Electromyography and Kinesiology*, 9 (6), 385-389.

126. Lintner D, Noonan TJ, Kibler WB. (2008) Injury patterns and biomechanics of the athlete's shoulder. *Clinics in Sports Medicine*, 27 (4), 527-551.

127. Ljungqvist A, Jenoure P, Engebretsen L, Alonso JM, Bahr R, Clough A, Thill C. (2009) The International Olympic Committee (IOC) Consensus Statement on periodic health evaluation of elite athletes March 2009. *British Journal of Sports Medicine*, 43 (9), 631-643.

128. Loeb GE, Yee WJ, Pratt CA, Chanaud CM, Richmond FJ. (1987) Cross-correlation of EMG reveals widespread synchronization of motor units during some slow movements in intact cats. *Journal of Neuroscience Methods*, 21, 239-249.

129. Lopez E, Saucedo F, Navarro, E, Martinez de Quel, O, de Antonio, R. (2007) Analysis of the fencing lunge flight phase in epee. Abstracts from 12th Annual Congress of the ECSS.

130. Louis N, Gorce P. (2010) Surface electromyography activity of upper limb muscle during wheelchair propulsion: Influence of wheelchair configuration. *Clinical Biomechanics*, 25 (9), 879-885.

131. Lowery MM, Stoykov NS, Kuiken TA. (2003) A simulation study to examine the use of cross-correlation as an estimate of surface EMG cross talk. *Journal of Applied Physiology*, 94, 1324-1334.

132. Luinge HJ, Veltink PH. (2005) Measuring orientation of human body segments

using miniature gyroscopes and accelerometers. *Medical & Biological Engineering & Computing,* 43 (2), 273-282.

133. Luinge HJ, Veltink PH, Baten CT. (1999) Estimating orientation with gyroscopes and accelerometers. *Technology and Health Care*, 7 (6), 455-459.

134. Mackey AH, Walt SE, Lobb GA, Stott NS. (2005) Reliability of upper and lower limb three-dimensional kinematics in children with hemiplegia. *Gait & Posture,* 22, 1-9.

135. Magarey ME, Jones MA. (2003) Dynamic evaluation and early management of altered motor control around the shoulder complex. *Manual Therapy*, 8 (4), 195-206.

136. Malone LA, Gervais PL, Steadward RD. (2002) Shooting mechanics related to player classification and free throw success in wheelchair basketball. *Journal of Rehabilitation Research and Development,* 39(6), 701-710.

137. Mantovani G, Ravaschio A, Piaggi, P, Landi A. (2010) Fine classification of complex motion pattern in fencing. *Procedia Engineering*, 2 (2), 3423-3428.

138. Martetschlger F, Rios D, Boykin RE, Giphart JE, de Waha A, Millett PJ. (2012) Coracoid impingement: current concepts. *Knee Surgery, Sports Traumatology, Arthroscopy*, 20, 2148-2155.

139. Mathiassen SE, Winkel J, Hägg GM. (1995) Normalization of surface EMG amplitude from the upper trapezius muscle in ergonomic studies—a review. *Journal of Electromyography and Kinesiology*, 5 (4), 197-226.

140. Mayagoitia RE, Nene AV, Veltink PH. (2002) Accelerometer and rate gyroscope measurement of kinematics: an inexpensive alternative to optical motion analysis systems. *Journal of Biomechanics*, 35, 537-542.

141. McCormack DAR, Reid DC, Steadward R, Syrotuik DG. (1991) Injury profiles in wheelchair athletes: results of a retrospective survey. *Clinical Journal of Sports Medicine,* 1, 35-40.

142. Meeuwisse WH. (1994) Assessing causation in sport injury: a multifactorial

model. *Clinical Journal of Sport Medicine*, *4*(3), 166-170.

143. Milne AD, Chess DG, Johnson JA, King GJW. (1996) Accuracy of an electromagnetic tracking device: a study of the optimal operating range and metal interference. *Journal of Biomechanics*, 29 (6), 791-793.

144. Morris N, Farnsworth M, Robertson DGE. (2011) Kinetic analyses of two fencing attacks-lunge and fleche. *Portuguese Journal of Sport Science*, *11*(2), 343-346.

145. Moyer J, Konin J. (1992) An overview of fencing injuries (abstract). *American Fencing*, 42 (4), 25.

146. Mulroy SJ, Farrokhi S, Newsam CJ, Perry J. (2004) Effects of spinal cord injury level on t he activity of shoulder muscles during wheelchair propulsion: an electromyographic study. *Archives of Physical Medicine and Rehabilitation*, 85 (6), 925-934.

147. Murgu AI. (2006) Fencing. *Physical Medicine and Rehabilitation Clinics of North America*, 17 (3), 725-736.

148. Naghavi H. (2002) Echocardiographic aspects of young elite fencers (abstracts). *British Journal of Sports Medicine,* 36, 6.

149. National Spinal Cord Injury Statistical Center. (2013) Spinal cord injury facts and figures at a glance. *The Journal of Spinal Cord Medicine*, 36 (1), 1-2.

150. Neto OP, Magini M. (2008) Electromyographic and kinematic characteristics of Kung Fu Yau-Man palm strike. *Journal of Electromyography and Kinesiology*, 18 (6), 1047-1052.

151. Newsam CJ, Rao SS, Mulroy SJ, Gronley JK, Bontrager EL, Perry J. (1999) Three dimensional upper extremity motion during manual wheelchair propulsion in men with different levels of spinal cord injury. *Gait & Posture,* 10, 223-232.

152. Nuesch C, Göpfert B, Fischer M, Frere J, Wirz D, Friederich N. (2008) Wavelet-EMG-analysis of the co-contraction of the leg muscles in fencing

during a flèche attack. *Forschungstag*, 69.

153. Nunome H, Doyo W, Sakurai S, Ikegmai Y, Yabe K. (2002) A kinematic study of the upper-limb motion of wheelchair basketball shooting in tetraplegic adults. *Journal of Rehabilitation Research and Development,* 39 (1), 63-71.

154. Nyland JN, Snouse SL, Anderson M, Kelly T, Sterling JC. (2002) Soft tissue injuries to USA Paralympians at the 1996 summer games. *Archives of Physical Medicine and Rehabilitation,* 81, 368-373.

155. Nystrom J, Lindwall O, Ceci R, Harmenberg J, Swedenhag J, Ekblom B. (1990) Physiological and morphological characteristics of world class fencers. *International Journal of Sports Medicine*, 11 (2), 136-139.

156. Pappas AM, Zawacki RM, Sullivan TJ. (1985) Biomechanics of baseball pitching. A preliminary report. *The American Journal of Sports Medicine*, 13 (4), 216-222.

157. Pearcy MJ, Gill JM, Hindle RJ, Johnson GR. (1987) measurement of human back movement in three dimensions by opto-electronic devices. *Clinical Biomechanics*, 2, 199-204.

158. Pepper M, Willick S. (2009) Maximizing physical activity in athletes with amputations. *Current Sports Medicine Reports*, 8 (6), 339-344.

159. Portney LG, Watkins MP. (2009) *Foundations of Clinical Research Applications to Practice*. 3rd edition. Upper Saddle River, NJ: Pearson & Prentice Hall.

160. Potten, YJM, Seelen HAM, Drukker J, Reulen JPH, Drost MR. (1999) Postural muscle responses in the spinal cord injured persons during forward reaching. *Ergonomics*, 42, 1200-1215.

161. Putnam CA. (1993) Sequential motions of body segments in striking and throwing skills: descriptions and explanations. *Journal of Biomechanics*, 26, 125-135.

162. Rab G, Petuskey K, Bagley A. (2002) A method for determination of upper extremity kinematics. *Gait & posture*, 15 (2), 113-119.

163. Ramakrishnan HK, Kadaba MP. (1991) On the estimation of joint kinematics during gait. *Journal of Biomechanics*, 24, 969–977.

164. Rau G, Disselhorst-Klug C, Schmidt R. (2000) Movement biomechanics goes upwards: from the leg to the arm. *Journal of Biomechanics,* 33, 1207-1216.

165. Reaz MBI, Hussain MS, Mohd-Yasin F. (2006) Techniques of EMG signal analysis: detection, processing, classification and applications. *Biological Procedures Online*, 8 (1), 11-35.

166. Reid M, Elliott B, Alderson J. (2007) Shoulder joint kinematics of the elite wheelchair tennis serve. *British Journal of Sports Medicine,* 41, 739-44.

167. Reid S, Elliott C, Alderson J, Lloyd D, Elliott B. (2010) Repeatability of upper limb kinematics for children with and without cerebral palsy. *Gait & posture*, 32 (1), 10-17.

168. Reynolds J, Stirk A, Thomas A, Geary F. Paralympic – Barcelona 1992. (1994) *British Journal of Sports Medicine,* 28 (1), 14-17.

169. Richter KJ, Hyman SC, Mushett CA. (1991) Injuries in world class cerebral palsy athletes of the 1988 South Korea Paralympics. *Journal of Osteopathic Sport Medicine,* 7, 15-18.

170. Rippetoe M. (2000) Strength and conditioning for fencing. *Strength & Conditioning Journal*, 22, 42-47.

171. Robertson EM, Miall RC. (1997) Multi-joint limbs permit a flexible response to unpredictable events. *Experimental Brain Research*, 117 (1), 148-152.

172. Roetenberg D, Slycke PJ, Veltink PH. (2007) Ambulatory position and orientation tracking fusing magnetic and inertial sensing. *IEEE. Transactions on Biomedical,* 54, 883-890.

173. Roi GS, Bianchedi D. (2008) The Science of Fencing. *Sports Medicine*, 38 (6), 465-481.

174. Roli GS, Fasci A. (1988) Survey of requests for medical assistance during fencing matches. *Italian Journal of Sports Traumatology.* 10 (1), 55-62.

175. Rota S, Rogowski I, Champely S, Hautier C. (2013) Reliability of EMG normalisation methods for upper-limb muscles. *Journal of Sports Sciences*, 31 (15), 1696-1704.

176. Roux E, Bouilland S, Bouttens D, Istas D, Godillon-Maquinghen AP, Lepoutre FX. (2000) *Evaluation of the kinematics of the shoulder and of the upper limb*. In: Third Conference of the International Shoulder Group.

177. Sabatini AM, Martelloni C, Scapellato S, Cavallo F. (2005) Assessment of walking features from foot inertial sensing. *Biomedical Engineering, IEEE Transactions on*, 52 (3), 486-494.

178. Saber-Sheikh K, Bryant EC, Glazzard C, Hamel A, Lee RY. (2010) Feasibility of using inertial sensors to assess human movement. *Manual Therapy*, 15 (1), 122-125.

179. Sabick MB, Kim YK, Torry MR, Keirns MA and Hawkins RJ. (2005) Biomechanics of the shoulder in youth baseball pitchers: implications for the development of proximal humeral epiphysiolysis and humeral retrotorsion. *American Journal of Sports Medicine*, 33, 1716-1722.

180. Sapega AA, Minkoff J, Nicholas JA, Valsamis M. (1977) Sport-specific performance factor profiling: fencing as a prototype. *The American Journal of Sports Medicine,* 6 (5), 232-235.

181. Sapega AA, Minkoff J, Valsamis M, Nicholas JA. (1984) Musculoskeletal performance testing and profiling of elite competitive fencers. *Clinics in Sports Medicine*, 3 (1), 231-244.

182. Seelen HA, Vuurman EF. (1991) Compensatory muscle activity for sitting posture during upper extremity task performance in paraplegic persons. *Scandinavian Journal of Rehabilitation Medicine*, 23 (2), 89-96.

183. Seelen HAM, Potten YJM, Adam JJ. (2001) Motor preparation in postural control in seated spinal cord injured people. *Ergonomics,* 44 (4), 457-472.

184. Seelen HAM, Potten YJM, Huson A, Spaans F, Reulen JPH. (1997) Impaired balance control in paraplegic subjects. *Journal of Electromyography and Kinesiology*, 7 (2), 149-160.

185. SENIAM. (1999) *European Recommendations for Surface Electromyography, deliverable of the SENIAM project*. In: Hermens, H.J.E.A. (Ed.), Roessingh Research and Development B.V.

186. Shiavi R, Green N. (1983) Ensemble averaging of locomotor electromyographic patterns using interpolation. *Medical and Biological Engineering and Computing*, 21 (5), 573-578.

187. Shum GL, Crosbie J, Lee R Y. (2005) Symptomatic and asymptomatic movement coordination of the lumbar spine and hip during an everyday activity. *Spine,* 30 (23), E697-E702.

188. Shum GLK, Crosbie J, Lee RYW. (2007) Movement coordination of the lumbar spine and hip during a picking up activity in low back pain subjects. *European Spine Journal*, 16, 749-758.

189. Shrout PE, Fleiss JL. (1979) Intraclass correlations: uses in assessing rater reliability. *Psychological Bulletin*, 86 (2), 420.

190. Sobiecka J. (2005) Injuries and ailments of the Polish participants of the 2000 Paralympic Games in Sydneys. *Biology of Sport*, 22 (4), 353.

191. Steinwender G, Saraph V, Scheiber S, Zwick EB, Uitz C, Hackl K. (2000) Intrasubject repeatability of gait analysis data in normal and spastic children. *Clinical Biomechanics,* 15, 134-139.

192. Stewart SL, Kopetka B. (2005) The kinematic determinants of speed in the fencing lunge (Abstract). *Journal of Sports Sciences,* 23, 105.

193. Stotts KM, Warren CG. (1982) Look at wheelchair sports and what's happening in northwest. *Model Systems SCI Digest*, 4, 13–18.

194. Suchanowski A, Borysiuk Z, Pakosz P. (2011) Electromyography Signal

Analysis of the Fencing Lunge by Magda Mroczkiewicz, the Leading World Female Competitor in Foil. *Baltic Journal of Health and Physical Activity*, 3 (3), 172-175.

195. Subramanian SK, Yamanaka J, Chilingaryan G, Levin MF. (2010) Validity of movement pattern kinematics as measures of arm motor impairment poststroke. *Stroke*, 41 (10), 2303-2308.

196. Suchanowski A, Borysiuk Z, Pakosz P. (2011) Electromyography Signal Analysis of the Fencing Lunge by Magda Mroczkiewicz, the Leading World Female Competitor in Foil. *Baltic Journal of Health and Physical Activity*, 3 (3), 172-175.

197. Szilagui T. (1993) Dynamic characterization of fencing lunge (Abstract). Presented at the International Society of Biomechanics XIVth Congress, Paris, July, 1314-1315.

198. Taylor D, Williams T. (1995) Sports injuries in athletes with disabilities: wheelchair racing. *Paraplegia,* 33, 296-299.

199. Thies SB, Tresadern P, Kenney L, Howard D, Goulermas JY, Smith C, Rigby J. (2007) Comparison of linear accelerations from three measurement systems during “reach & grasp”. *Medical Engineering & Physics*, 29, 967-972.

200. Toyoshima S, Hoshikawa T, Miyashita M, Oguri T. (1974) Contribution of the body parts to throwing performance. *International Series on Sports Sciences*, 1, 169-74.

201. Tsolakis CH, Katsikas CH. (2006) Long term effects of a combined physical conditioning and fencing training program on neuromuscular performance in elite fencers. *International Journal of Fitness*, 2 (1), 35-42.

202. Turner A, Miller S, Stewart P, Cree J, Ingram R, Dimitriou L, Kilduff L. (2013) Strength and conditioning for fencing. *Strength & Conditioning Journal*, 35 (1), 1-9.

203. Uzun S, Pourmoghaddam A, Hieronymus M, Thrasher TA. (2012) Evaluation of muscle fatigue of wheelchair basketball players with spinal cord injury using

recurrence quantification analysis of surface EMG. *European journal of applied physiology*, 112 (11), 3847-3857.

204. van Mechelen W, Hlobil H, Kemper HC. (1992) Incidence, severity, aetiology and prevention of sports injuries. A review of concepts. *Sports Medicine,* 14, 82-99.

205. VencesBrito AM, Rodrigues Ferreira MA, Cortes N, Fernandes O, Pezarat-Correia P. (2011) Kinematic and electromyographic analyses of a karate punch. *Journal of Electromyography and Kinesiology*, 21 (6), 1023-1029.

206. Wang L, Hu W, Tan, T. (2003) Recent developments in human motion analysis. *Pattern Recognition*, 36 (3), 585-601.

207. Wang X, Maurin M, Mazet F, Maia ND, Voinot K, Verriest JP, Fayet M. (1998) 3-D modeling of the motion range of axial rotation of the upper arm. *Journal of Biomechanics*, 31, 899–908.

208. Webborn AD. (1999) Fifty years of competitive sport for athletes with disabilities: 1948-1998. *British Journal of Sports Medicine*, 33 (2), 138.

209. Webborn AD, Turner HM. (2000) The aetiology of shoulder pain in elite Paralympic wheelchair athletes- the shoulder or cervical spine? *5th Paralympic Scientific Congress. Sydney, Australia: International Paralympic Committee,* 59.

210. Webborn N. (2012) Lifetime injury prevention: the sport profile model. *British Journal of Sports Medicine*, 46 (3), 193-197.

211. Webborn N, Van de Vliet P. (2012). Paralympic medicine. *The Lancet*, 380 (9836), 65-71.

212. Webborn N, Willick S, Emery CA. (2012) The injury experience at the 2010 winter paralympic games. *Clinical Journal of Sport Medicine*, 22 (1), 3-9.

213. Williams JM, Haq I, Lee RY. (2013) An investigation into the onset, pattern, and effects of pain relief on lumbar extensor electromyography in people with acute and chronic low back pain. *Journal of Manipulative and Physiological*

Therapeutics, 36, 91-100.

214. Williams LRT, Walmsley A. (2000) Response timing and muscular coordination in fencing: a comparison of elite and novice fencers. *Journal of Science and Medicine in Sport*, 3 (4), 460-475.

215. Willick SE, Webborn N, Emery C, Blauwet CA, Pit-Grosheide P, Stomphorst J, Schwellnus M. (2013) The epidemiology of injuries at the London 2012 Paralympic Games. *British Journal of Sports Medicine*, 47 (7), 426-432.

216. Wilson PE, Washington RL. (1993) Pediatric wheelchair athletic: sports injuries and prevention. *Paraplegia*; 31, 330-337.

217. Winter DA. (1990) *Biomechanics and motor control of human movement*. 2nd edition. John Willey & Sons: New York.

218. Winter DA, Yack HJ. (1987) EMG profiles during normal human walking: stride-to-stride and inter-subject variability. *Electroencephalography and Clinical Neurophysiology*, 67 (5), 402-411.

219. World report on disability. Geneva, World Health Organization 2011.

220. Wren TAL, Do KP, Rethlefsen SA, Healy B. (2006) Cross-correlation as a method for comparing dynamic electromyography signals during gait. *Journal of Biomechanics*, 39, 2714-2718.

221. Woltring HJ. (1994) 3-D attitude representation of human joints: a standardization proposal. *Journal of Biomechanics,* 27, 1399-1414.

222. Woodward M. (2005) Epidemiology: Study Design and Data Analysis. 2nd ed. Boca Raton, FL: Chapman & Hall/CRC.

223. Worsley P, Warner M, Mottram S, Gadola S, Veeger HEJ, Hermens H, Morrissey D, Little P, Cooper C, Carr A, Stokes M. (2013) Motor control retraining exercises for shoulder impingement: effects on f unction, muscle activation, and biomechanics in young adults. *Journal of Shoulder and Elbow Surgery*, 22 (4), e11-19.

224. Wu G, Cavanagh PR. (1995) ISB recommendations for standardization in the reporting of kinematic data. *Journal of Biomechanics,* 28, 1257–1261.

225. Wu G, van der Helm FCT, Veeger HEJ, Makhsous M, Van Roy P, Anglin C, Nagels J, Karduna AR, McQuade K, Wang X, Werner FW, Buchholz B. (2005) ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion – Part II: shoulder, elbow, wrist and hand. *Journal of Biomechanics,* 38 (1), 981-992.

226. van Mechelen W, Hlobil H, Kemper HC. (1992) Incidence, severity, aetiology and prevention of sports injuries. *Sports Medicine*, 14 (2), 82-99.

227. Vanlandewijck Y, Thompson WR. (Eds.). (2011) *The Paralympic Athlete: Handbook of Sports Medicine and Science*. Wiley-Blackwell.

228. Vicon Upper Limb Model Product Guide, Version 1.0. (2007) Oxford, UK: Oxford Metrics Ltd.

229. Yildirim NU, Comert E, Ozengin N. (2010) Shoulder pain: a comparison of wheelchair basketball players with trunk control and without trunk control. *Journal of Back and Musculoskeletal Rehabilitation,* 23 (2), 55-61.

230. Zazulak BT, Hewett T E, Reeves NP, Goldberg B, Cholewicki J. (2007) Deficits in neuromuscular control of the trunk predict knee injury risk a prospective biomechanical-epidemiologic study. *The American Journal of Sports Medicine*, 35 (7), 1123-1130.

231. Zemper E, Dick R. (2007) Epidemiology of athletic injuries. In: McKeag D., Moeller J., Zemper E, eds. *Primary Care Sports Medicine (2nd Edition).* Philadelphia, PA: Williams & Wilkins, 63-73.

232. Zijdewind I, Kernell D, Kukulka CG. (1995) Spatial differences in fatigue-associated electromyographic behaviour of the human first dorsal interosseus muscle. *The Journal of Physiology*, 483 (Pt 2), 499-509.

233. Zhou H, Hu H. (2007) Inertial sensors for motion detection of human upper limbs. *Sensor Review*, 27, 151.

234. Zhou H, Hu H. (2010) Reducing drifts in the inertial measurements of wrist and elbow positions. Instrumentation and Measurement, *IEEE Transactions on Biomedical,* 59 (3), 575-585.

235. Zhou H, Hu H, Tao Y. (2006) Inertial measurements of upper limb motion. *Medical Biological Engineering & Computing,* 44, 479-487.

236. Zhou H, Stone T, Hu H, Harris N. (2008) Use of multiple wearable inertial sensors in upper limb motion tracking. *Medical Biological Engineering & Computing,* 30, 123-133.