The Hong Kong Polytechnic University

Department of Rehabilitation Sciences

Effect of Resistance Training on

Active and Passive Patellar Stabilizers

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A thesis submitted in partial fulfillment of the requirements

for the degree of Doctor of Philosophy

2008

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Certification of Originality

I, Wong Yiu Ming, declare that this thesis, submitted in partial fulfillment of the degree of Doctor of Philosophy, at the Department of Rehabilitation Sciences, The Hong Kong Polytechnic University, is wholly my own work unless otherwise referenced or acknowledged. This thesis has not been submitted for qualifications at any other academic institute.

Signed:

Dated:

Abstract

Patellofemoral pain syndrome (PFPS), which affects approximately 25% of the population at some stage in their lives, could be caused by an unstable patellofemoral joint leading to excessive lateral patellar tracking in postural and functional loadings, which eventually strains the retro-patellar subchondral bone and triggers the localized anterior knee pain. Although various medical and physical treatments have been prescribed for people with PFPS, literature on prevention of PFPS is lacking. Thus, the overall purpose of this thesis is to explore effects of physical conditioning on patellar stabilizers among previously untrained adults, in terms of biomechanical, morphological and functional changes of the passive and active patellar stabilizing structures.

Five experiments were conducted to develop and validate the outcome measurements and the main study was a prospective investigation with 48 volunteers randomly allocated in three groups for 8 weeks, namely 1) weight training for muscular hypertrophy, 2) weight training for muscle strength, and 3) non-exercising control group. The subjects in the intervention groups were trained 3 times a week and all subjects were measured twice with 8 weeks apart for their cross-section of vastus medialis obliquus (VMO), patellar tilt angle and mobility, knee extension force, knee joint position sense and electromyographic (EMG) characteristics of the distal quadriceps.

After the 8-week trainings, the subjects in both training groups had comparable and significant increases in the VMO size, passive patellar stability, knee extensor force, knee joint position sense and EMG onset and amplitude of vasti muscles, whilst the control group showed no significant change throughout the 8 weeks of study. The findings of this thesis form a basis for the establishment of a PFPS-prevention exercise program and enhance our understanding of the bio-kinesiology of the patellofemoral joint under the influence of systematic physical conditioning.

Publications Arising from the Thesis

Journal articles:

1. Wong YM, Ng GYF (2005) The double peak-to-peak analysis for determining EMG onset of muscle contraction. *Electromyography & Clinical Neurophysiology*, 45(5): 267-271

2. Wong YM, Ng GYF (2006) Surface electrode placement affects the EMG recordings of the quadriceps muscles. *Physical Therapy in Sport*, 7(3):122-127

3. Wong YM, Ng GYF (2008) The relationships between the geometrical features of patellofemoral joint and patellar mobility in able-bodied subjects. *American Journal of Physical Medicine & Rehabilitation*, 87(2):134-138

4. Wong YM, Chan ST, Tang KW, Ng GYF. Comparing different modes of weight training programs on patellar stabilization. *Journal of Athletic Training*, Pending on revision

Conference abstracts:

1. Wong YM, Ng GYF (2005) Quantitative assessment for force-displacement relationship of patellofemoral Joint. *The 2nd Asia Pacific Conference on Biomechanics*. Taipei, Taiwan, November 23-25, 2005

2. Wong YM, Ng GYF (2006) Relationship between patellar mobility and MRI knee images. *The 3rd Post-graduate Student Seminar, Theme: Musculoskeletal Sciences in Rehabilitation*. Hong Kong Polytechnic University, Hong Kong, March 18, 2006

3. Wong YM, Ng G (2006) Correlation between MRI and mechanical measurements for patellar stability. *American Society of Biomechanics Annual Meeting*. Blacksburg, VA, USA, September 7-9, 2006

4. Wong YM, Ng GYF (2007) Reliability of hand-free ultrasound measurement for vastus medialis obliquus. *American Society of Biomechanics Annual Meeting*. San Francisco, CA, USA, August 23-25, 2007

Acknowledgements

This thesis could not have been completed without the valuable advice and patient guidance of my supervisor Professor Gabriel Ng. My sincere appreciation is extended to the voluntary subjects who participated in my studies.

I want to thank Dr. Chris Powers who gave me the opportunity to be involved in his research work on patellofemoral joint during my 4-month attachment at the University of Southern California.

My thanks also go to Dr. Phoebe Chan, Dr. K.W. Tang, Dr. Raymond Chung, Dr. Bruno Herbelin, Dr. C.Y. Tang, Mr. Kyle Smith, Mr. Peggo Lam, Mr. S.C. Siu, Mr. Man Cheung and all members of Professor Gabriel Ng's research team for their advice, comments and assistance in the progress of my study. I would like to thank the Hong Kong Polytechnic University for granting me the postgraduate traveling grant for my participating in overseas attachment and conferences.

Lastly, my wife receives my deepest gratitude for her many years of support and encouragement.

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List of Abbreviations

AC	=	Alternative current
ANOVA	=	Analysis of variance
CI	=	Confidence intervals
CSA	=	Cross sectional area
CT	=	Computed tomography
DC	=	Direct current
df	=	Degrees of freedom
deg	=	Degree
DP-P	=	Double peak to peak
EMG	=	Electromyography
ICC	=	Intraclass correlation coefficient
LPFA	=	Lateral patellofemoral angle
mm	=	Millimeter
MMG	=	Mechanomyography
MPFL	=	Medial patellofemoral ligament
MPML	=	Medial patellomeniscal ligament
MRI	=	Magnetic resonance image
MVC	=	Maximal voluntary contraction
PTA	=	Patellar tilt angle
RMS	=	Root mean square
RF	=	Rectus femoris
s.d., SD	=	Standard deviation
TE	=	Time to echo
TI	=	Time of inversion
TL/PL	=	Patellar tendon length / patellar length
TR	=	Time to repeat
VI	=	Vastus intermedius
VL	=	Vastus lateralis
VMO	=	Vastus medialis obliquus

Chapter 1

Introduction

1.1. Background

Patellofemoral Pain Syndrome (PFPS) describes the condition of pain over the anterior aspect of the knee that is exacerbated by functional or physical activities without other identifiable causative pathologies (Cutbill et al 1997; LaBotz 2004). The PFPS affects approximately 25% of the population at some stage of their lives (McConnell 1996) with females being more vulnerable than males (Almeida et al 1999); the pain can be chronic and even lasts for years (Leppälä et al 1998; Jensen et al 2007; Souza & Powers 2008). The PFPS may be caused by a combination of intrinsic and extrinsic factors. Most of the intrinsic factors relate to an excessive lateral patellar tracking that leads to excessive strain on the retro-patellar subchondral bone (Thomee et al 1999; Draper et al 2006), and these include an imbalance of vasti muscles (Boling et al 2006), increased Q-angle (LaBella 2004), hyper-mobile patella (Witvrouw et al 2000), or tightness of patellar retinaculum (Fulkerson 1983). Extrinsic factors commonly contributed to PFPS include poor form of exercise (Powers 2003), excessive duration or frequency of physical activities (Dixit et al 2007). An early study by Fox (1975) reported that 40% of his orthopedic patients without knee pain had imbalanced muscle development of vastus medialis obliquus (VMO) to vastus lateralis (VL) and they were prone to develop PFPS.

1.2. Measurable quantities and predisposing factors of PFPS

Although the pathology underlying PFPS is believed to be of multi-faceted and the exact mechanisms are not yet fully understood, certain measurable quantities of PFPS have been identified in retrospective studies. Kujala et al (1993) reported a positive correlation between the lateral patellar tilt by magnetic resonance imaging (MRI) measurement in patients with anterior knee pain and their subjective symptoms. Pookarnjanamorakot et al (1998) analyzed the degree of lateral patella tilt with MRI and its correlation with the symptoms of anterior knee pain, and found that the average patella tilt angle for subjects without anterior knee pain was 12.8° (SD = 8.4); whereas it was 6.3° (SD = 3.9) for people with anterior knee pain.

Souza & Gross (1991), Boucher et al (1992), and Makhsous et al (2004) reported that subjects with PFPS had reduced VMO muscle recruitment in terms of decreased electromyographic (EMG) amplitude of VMO/VL ratio as compared to asymptomatic controls in static knee extension. On the temporal aspect, Cowan et al (2001) monitored the EMG onset time difference between VMO and VL in subjects with PFPS and able-bodied controls during stair-stepping. They found that the onset of VL occurred before that of VMO in both the step up/down phases in subjects with PFPS, but no such difference was found for the controls. In addition, Baker et al (2002) reported the decreased knee joint position sense in people with PFPS when compared to asymptomatic subjects. Callaghan & Oldham (2004) found an 18.4% difference in the isokinetic knee extension peak torque between affected and unaffected legs of patients with PFPS, whist the controls had only a difference of 7.6% between their dominant and non-dominant legs.

Witvrouw et al (2000) conducted a 2-year prospective study identifying the predisposing factors of PFPS among 282 students aged 17-21. They reported that a delayed VMO reflex response time, under-conditioned vertical jump height, inflexible quadriceps and a hypermobile patella had significant correlations with the incidence of patellofemoral pain in that population.

1.3. Motivation and statement of the problem

In the treatment of PFPS, surgery may solve the problems of structural abnormalities such as tight lateral retinaculum (Fu & Maday 1992), whereas physical therapy may correct the VMO-VL muscle imbalances (Cowan 2002b; Boling et al 2006). Most clinical studies (Cowan et al 2002b; Hazneci et al 2005; O'Sullivan & Popelas 2005; Yip & Ng 2006; Boling et al 2006) have focused on the rehabilitation exercises when PFPS was already developed but preventative exercises for PFPS have not been well investigated.

Ingersoll & Knight (1991) showed that asymptomatic subjects undergoing only 3 weeks of resistive or EMG biofeedback quadriceps training were able to change their patellar positions under the Merchant's view of X-ray examination. They hypothesized that the shortened knee extensor muscles changed the position and might therefore promote a more favorable patellar tracking, which could decrease the predisposition to lateral patellar instability.

However, resistive training does not only strengthen the skeletal muscles, but the adaptations occur in the adjacent tendons, ligaments and fasciae. The enhanced connective tissue network could develop to withstand the greater tensile loading of the stronger muscles (Wren et al 2000). As the patellofemoral joint is a freely movable joint stabilized by muscles and connective tissues over the anterior knee, resistive weight training programs could have a holistic effect on the physiological functioning of this joint. As there is little information in the literature on the aforesaid resistive training effect on patellofemoral joint, the present thesis is conducted to investigate the potential role of resistive weight training that may impact on the quadriceps muscle and the associated connective tissues for the functioning of the patellofemoral joint. Cognizant of the strategic importance of the VMO and the medial patellofemoral ligament being the primary active and passive stabilizers of the patellofemoral joint (Owings & Grabiner 2002; Steensen et al 2004), this thesis focuses on the physiological functioning of these structures in maintaining the normal kinematic and kinetic integrity of the patella.

1.4. Anatomy of patellofemoral joint

In order to understand the mechanical functioning of the patellofemoral joint, it is necessary to review anatomical structures of the lower extremity that constitute to the joint. The following is a brief review of the anatomy of the patellofemoral joint.

1.4.1. Skeletal and cartilaginous structure

The patellofemoral joint is a synovial joint with considerable mobility. An able-bodied person can have their patella passively mobilized in any direction over 1 cm along the frontal plane while the knee is extended and the quadriceps is relaxed (Fox 1975). The patella, as the central structure of the patellofemoral joint, is the largest sesamoid bone in the human body. Being triangular in shape and embedded within the quadriceps tendon, 75% of the posterior surface of patella is covered by hyaline cartilage, divided into two asymmetrical facets by a vertical ridge. The area of the lateral facet is usually broader and longer than the medial facet with an average width ratio of 1.4 to 1 (Minkoff & Fein 1989) (fig 1-1). Also, both the lateral and medial facets on the posterior patella can be divided horizontally into superior, middle and inferior regions (fig 1-2). Draper et al (2006) investigated asymptomatic subjects with MRI and reported that the average thickness of the middle region was 4.2 mm; while the superior and inferior region was 3.3 mm and 2.9 mm, respectively.

As a synovial joint, the patella articulates with the trochlear sulcus (also known as femoral sulcus) of the distal femur. The trochlear sulcus is concave in shape and formed by the lateral and medial femoral facets. The sulcus angle is normally less than 150° and the lateral condyle is more prominent anteriorly than the medial condyle (Carrillon et al 2000; McNally et al 2000) (figs 1-3). People with shallow trochlear sulcus and hypoplastic lateral condyle are prone to develop patellar instability (Carrillon et al 2000; Van Huyssteen et al 2006) (fig 1-4). With MR imaging to 11 able-bodied males aged 11-42 years, Van Huyssteen et al (2006) found that the average cartilage thickness in the lateral and medial condyles was 3.1 mm and 2.7 mm correspondingly.

The anatomical features of the patellofemoral joint, including the relatively thicker cartilage of the lateral trochlear sulcus and the larger lateral facet of the patella, is correlated to experiments that indicated that the lateral patellofemoral contact force is greater than the medial compartment during postural and functional loadings. Heegaard et al (1995) measured the magnitude of lateral and medial patellofemoral joint contact forces using finite element modeling and they found that the Q angle and the knee flexion angle affected the mediolateral distribution of the joint contact force, as an increase in these angles was associated with an increase in the lateral patellofemoral contact force.

Eckstein et al (1998) inspected the thickness of the calcified cartilage of human cadaveric human patellae, and found that the calcified cartilage located on the lateral facet of patella was significantly thicker than that of the medial side. The increased thickness of the calcified cartilage (subchondral mineralized tissue) was believed to be due to the long-term distribution of stress on the articular joint surface (Shimizu et al 1993).

Besier et al (2005) used an EMG-driven finite element modeling to estimate the patellofemoral stress during 0-60° squatting. They also found that the contact stress was higher on the lateral patellar cartilage and the corresponding lateral femoral cartilage as compared to that of the medial side.

As a standard feature in all synovial joints, there is a joint capsule linking the patella to the femur and tibia. The capsule is thin and loose in order to accommodate the large range of knee flexion that is approximately 0-145°, and it has little contribution to the stability of the patellofmoral joint (Reider et al 1981). The stability and mobility of the patella is mostly determined by the passive and active patellar stabilizers (Fulkerson 2004).

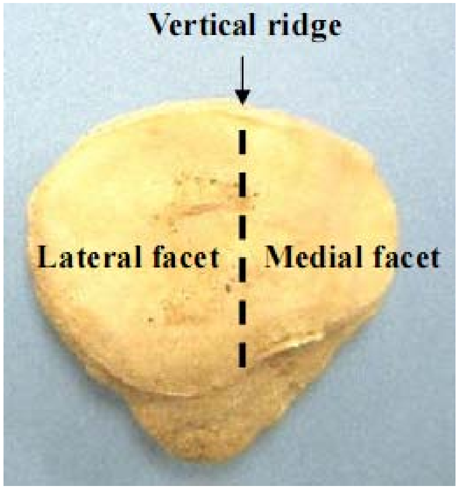


Figure 1-1:
The posterior view of a typical left patella*

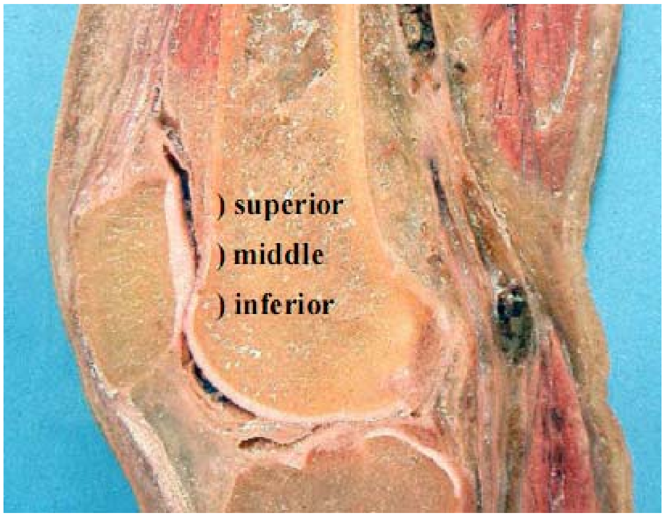


Figure 1-2:
Sagittal view of the mid-patella. It shows different regions of the patellar cartilage with distinguishable thickness in vitro (plastinated specimen)*

* = Anatomical specimen provided by the anatomy lab, Hong Kong Polytechnic University with permission.

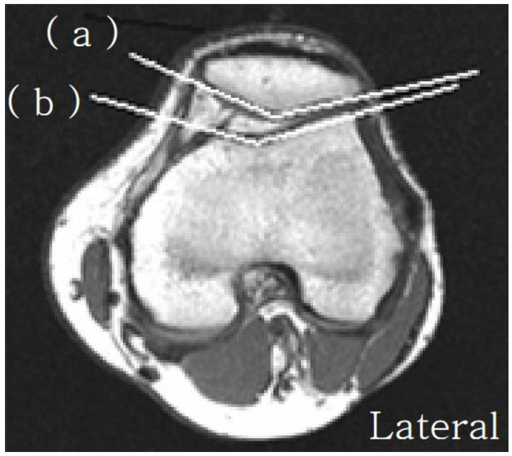


Figure 1-3: MRI transverse view at the mid patellar level while the knee extended.
(a) Line drawn along the lateral and medial facet of the patella (patellar angle)
(b) Sulcus angle measured at the transverse mid-patellar level

The patellofemoral joint cavity is within the line (a) and (b). The sulcus angle is usually slightly larger than that of the patellar facets, thus there is a space for the patellar tilting to occur (Fujikawa et al 1983).

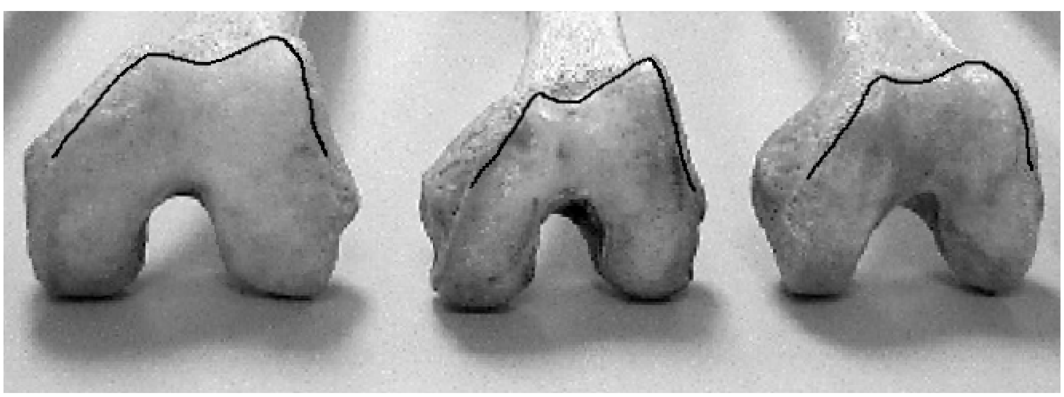


Figure 1-4: Left femurs with the articular surface partially outlined in inferior view. The femur at the right had a relatively flat trochear sulcus, while the middle one had a relatively prominent lateral condyle*

* = Anatomical specimen provided by the anatomy lab, Hong Kong Polytechnic University with permission.

1.4.2. Passive patellar stabilizers

Patellar tendon

The patella is connected to the tibia inferiorly through the patellar tendon (also known as the patellar ligament), which is broad-shaped of about 4-5 cm long and it only allows minimal amount of elongation due to contraction of the quadriceps. Hansen et al (2006) tested 12 able-bodied males (30 ± 7 years) by measuring their patellar tendon during voluntary maximal isometric knee extension at 90° knee flexion (open kinetic chain action). They found that elongation of the patellar tendon was approximately 4 mm as measured by the B-mode ultrasound. More recently, Defrate et al (2007) reported that the elongation of patellar tendon increased with loaded dynamic knee flexion. By using a real-time fluoroscopy to view the patellar tendon, they found the tendon to have elongated by approximately 6 mm in the 0-110° single leg lunge (closed kinetic chain action) among eleven asymptomatic subjects of mixed gender of mean age at 31 ± 8 years.

In addition, the length of patellar tendon can also affect patellar stability. A long patellar tendon or a high-riding patella is clinically known as “patellar alta”. It is determined by the PT/PL ratio. The ratio refers to the proportion calculated by the length of the patellar tendon divided by the length of the patellar bone (fig 1-5) and this ratio has been correlated to patellar subluxation or dislocation. If the ratio is over 1.3 as measured on the MRI sagittal view with knee extended (Miller et al 1996), or over 1.25 on the lateral x-ray with flexed knee at 30° (Insall et al 1972), this would be regarded as clinically long patellar tendon. Ward et al (2007) have demonstrated that the patellar alta can negatively affect patellar stability because of the relatively reduced patellofemoral joint contact area.

Figure 1-5: PT/PL ratio. The ratio is formed by the patellar tendon length divided by the diagonal patellar length at the MRI sagittal image. This example illustrates that the PT/PL ratio is 0.98 which is within normal range.

Peripatellar retinacula

The peripatellar retinacula have deep and superficial layers (Fulkerson & Gossling 1980; Fulkerson 1990; Nonweiler & DeLee 1994; Nomura et al 1995; Ford & Post 1997; Powers et al 2006), and researchers (Warren & Marshall 1979; Ruiz & Erickson 1994) even described the retinacula as consisting of three layers as they included the capsulosynovial component as the deepest layer. The superficial layer of the retinacula is attached to the patella and patellar tendon and extends to the fascia of sartorius medially, and the iliotibial band laterally (Fulkerson & Gossling 1980; Powers et al 2006). While it had been reported that the superficial layer is not functionally important in stabilizing the patella (Desio et al 1998), recently an *in-vitro* study (Powers et al 2006) found that if the retinacula was removed, the tensile loading of the patellar tendon increased 9.6 -16.6% while the quadriceps was stimulated to contract at 0-60° of knee flexion angles. Since both the superficial and deep layer of the retinacula were removed in the study, the role of the superficial patellar retinacula in the knee extension force transmission or patellar stability has not been fully understood.

In the deep layer of the medial retinacula, medial patellofemoral ligament (MPFL) has been reported to be the major passive restraint to lateral patellar displacement (Conlan et al 1993; Desio et al 1998). The MPFL is a fan-shaped fascial band of approximately 50 mm that originates from the adductor tubercle and extends to the upper two third of the medial patellar border (Conlan et al 1993; Tuxoe et al 2002). The ligament is tightly adherent to the overlying VMO aponeurosis (Conlan et al 1993). Several researchers (Bencardino et al 2000; Panagiotopoulos et al 2006) described the anatomical adhesion as the MPFL-VMO complex. Sallay et al (1996) also found that both the MPFL and the VMO are usually ruptured simultaneously during acute lateral patellar dislocation.

Locating inferior to the MPFL, the patellotibial and patellomeniscal ligaments act as secondary medial patellar restraints (fig 1-6). Conlan et al (1993) studied the deep medial retinaculum of 25 cadaveric knees with biomechanical testing technique and found that the MPFL contributed 53 ± 15% of the resistance to lateral translation at full knee extension. The medial patellomeniscal ligament and the medial patellotibial ligament contributed 22 ± 9.5% and 5 ± 5.9% of resistance to the lateral displacing force respectively.

In a study with 9 cadaveric knees from donors aged 43-70 years, Desio et al (1998) reported that the MPFL contributed 60 ± 13% of the total resistant force when the patella was translated at 20° of knee flexion; the patellomeniscal ligament and the patellotibial ligament contributed 13 ± 10% and 3 ± 3% respectively.

Since medial patellar subluxation or dislocation is clinically uncommon, the biomechanical property of the deep layer on the lateral retinaculum is less well studied than that of the medial side (Nonweiler & DeLee 1994). Anatomically, the deep layer consists of the lateral patellofemoral ligament (also known as epicondylopatellar band), the transverse ligament and the lateral patellotibial band with the latter two originating from the distal iliotibial tract to the lateral patellar border (fig 1-6). The transverse ligament may have a role as the primary lateral passive patellar stabilizer (Fulkerson & Gossling 1980), but this has yet to be proved by further research. A recent study with MRI examination on asymptomatic subjects showed that the iliotibial tract located against the lateral femoral epicondyle at 30° of knee flexion, and moved laterally during knee extension (Fairclough et al 2006). Also, the gluteus maximus and the tensor fasciae latae insert into the iliotibial tract, thus these muscles' action may indirectly affect the patellar kinematics.

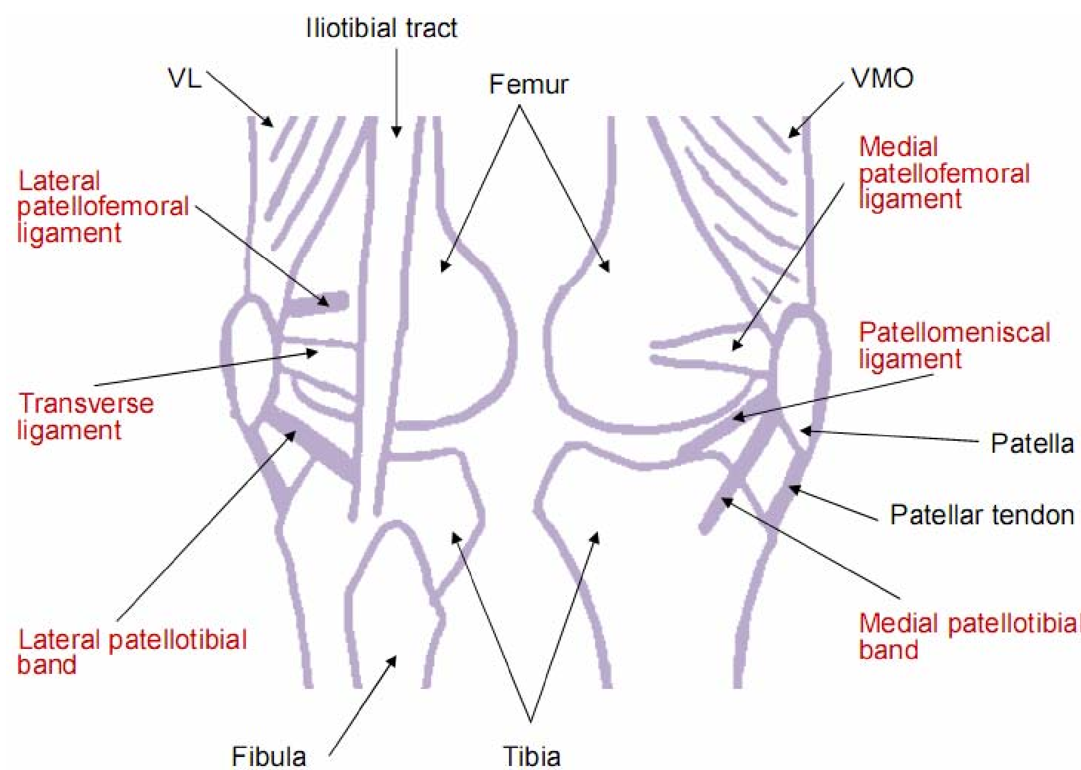


Figure 1-6: Components of the deep peripatellar retinacula. The components labeled in red of a left knee shown in lateral and medial view (modified from Fulkerson & Gossling 1980; Powers et al 2006; Panagiotopoulos et al 2006).

The main function of the medial and lateral retinacula is to control the optimal patellar tracking within the femoral groove and to equalize compression force between the medial and lateral facets of the patella (Ishibashi et al 2002; Farahmand et al 2004).

1.4.3. Active patellar stabilizers

The quadriceps is composed of four distinctive muscles, namely rectus femoris (RF), vastus intermedius (VI), vastus medialis (VM) and vastus lateralis (VL). The muscles are innervated by the femoral nerve that comes from L2-4 of posterior division of lumbar plexus. They have a common insertion at the base of patella and via the patellar tendon to the tibial tuberosity (Van De Graaff 1992). When acting as a group, contraction of the quadriceps muscles would lead to pulling the patella superiorly. The quadriceps angle (also known as Q-angle) is formed between the line of orientation of the quadriceps and the direction of the patella tendon. The normal value of the Q-angle is usually less than 15° (Herrington & Nester 2004). The resultant vector between the quadriceps and the patellar tendon tends to displace the patella laterally during active knee extensions (Hungerford & Barry 1979).

If the four muscles in the quadriceps are viewed separately, the RF and VI are located anterior to femur, and they primarily produce knee extension on the sagittal plane. The VL and VM locate on each side of the quadriceps, produce knee extension synergistically on the sagittal plane but antagonistically on the frontal plane for medial/lateral patellar movement. Therefore, they have been regarded as the primary active patellar stabilizers, particularly the distal muscle fibers (Cowan et al 2001; Owings & Grabiner 2002; Fulkerson 2004).

The VL originates from the greater trochanter and the lateral linea aspera of femur, and inserts to the superolateral edge of patella. Its muscle fibers angulate towards the shaft of femur at 15-20 degrees laterally (Weinstabl et al 1989; Van De Graaff 1992). The VM originates from medial linea aspera of proximal femur and inserts to the superomedial edge of patella (Van De Graaff 1992). Both the origin and insertion of VL are located slightly superiorly than that of the VM.

According to the classic study of Lieb & Perry (1968), it was demonstrated that the VM can further be divided into vastus medialis obliquus (VMO) and vastus medialis longus (VML). The muscle fiber orientation of the VML is approximately 15°, while that of the VMO is about 55° with respect to the shaft of femur. Weinstabl et al (1989) and Thiranagama (1990) reported that the VMO and VML can be defined by means of the two separate nerve branches and a fascial plane between the muscles. Also, Conlan et al (1993) reported that the VMO has an additional origin from the adductor magnus tendon proximal to the adductor tubercle.

However, there are different views on the anatomy of the VM muscle. Nozic et al (1997) regarded the separate VMO innervation as exceptional and it only existed in one out of 50 people; and therefore the study did not support the notion that VMO and VML functioning as two separate muscles. Additionally, Hubbard et al (1997) had dissected 374 adult cadaveric lower extremities and did not support that the VMO is an anatomically separate structure from the VML, but they confirmed a difference in fiber orientation between the proximal and distal VM muscle.

Both the VMO and VL are active throughout knee extension range of motion as revealed by a fine-wire electromyographic examination (Basmajian & De Luca 1985b). For the open kinetic chain knee extension, Lieb and Perry (1968) tested six cadaveric legs and concluded that the VMO keeps the patella centered on the trochlear sulcus by overcoming the tracking effect of VL during knee extension (0-90° knee flexion). The study by Goh et al (1995) with six cadaveric leg specimens study indicated that the absence of VMO tension causes the patella to displace laterally and increases the load on lateral patellar facet throughout the range of knee extension from 20-90° knee flexion. For closed kinetic chain knee extension, Ng et al (1998) measured five human knee specimens' patellofemoral joint pressure with different VMO tensions; they found that there was a general increase in the lateral patellofemoral contact pressure with a decrease in VMO tension at 30° and 90° knee flexion. Besier et al (2005) used the finite element model analysis and found that if the VMO muscle strength increased by 20%, the lateral patellar cartilage shear stress shifted medially without significant rise of the total patellar cartilage shear stress in squatting. Thus, the balanced synergism between VMO and VL should play an important role in controlling patellofemoral joint contact pressure distribution during movements.

1.4.4. Kinematics of patella

Throughout the entire range of knee flexion, the patella slides 5-7 cm over the trochlear groove of the distal femur (Hehne 1990). The path of the patellar sliding is not a straight line, but is determined by a complex interaction between the passive and active patellar stabilizers and the geometrical shape of the patellofemoral articular surface (Fulkerson 2004).

The normal patellar kinematics has been investigated in numerous studies. Since different measurement methods were used both in *vitro* and in *vivo*, many different patellar tracking patterns have been described (Katchburian et al 2003). The most universally agreed patellar tracking is the course of concave lateral curve along the sagittal plane during dynamic knee flexion (fig 1-7).

At full knee extension, the inferior patella rests in contact with the supratrochlear fat pad. As the knee flexes, the patella shifts medially and once it enters the trochear sulcus at 20° to 30° of flexion, it begins to shift laterally, thus the first articular contact is made between the patella and the trochlear sulcus. The articular contact area increases with further knee flexion. Between 40° to 90° of knee flexion, the patella continues to shift laterally and sinks deeper into the trochlear sulcus (fig 1-8), then the lateral facet of the trochlear sulcus functions as an inclined structure to prevent further lateral patellar translation. From 100° to full knee flexion, the medial and lateral areas of the upper patella make contact with the respective femoral condyles (Hungerford & Barry 1979; Reider et al 1981; Fujikawa et al 1983; Sheehan et al 1999; Katchburian et al 2003).

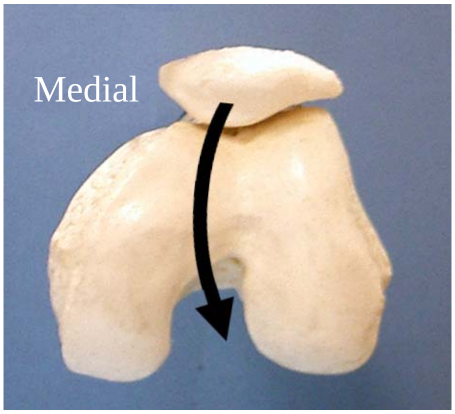


Figure 1-7: Patellar tracking path. The path follows a concave lateral curve from full extension to full flexion of knee. Inferior view of a left femur.

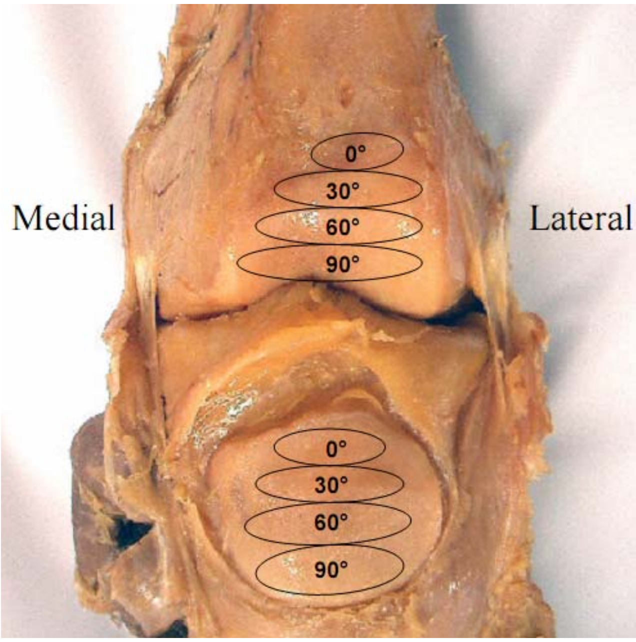


Figure 1-8: Contact areas between the patella and the distal femur during various knee flexion angles. Anterior view of a left knee with patella flipped downward* (adopted from Hinterwimmer et al 2004; Draper et al 2006).

* = Anatomical specimen provided by the anatomy lab, Hong Kong Polytechnic University with permission.

1.4.5. Summary on anatomy of patellofemoral joint

The patellofemoral joint as a freely movable joint, consists of multiple elements that act in a complex synergistic unit. Although anatomical difference exists among individuals, the joint, in general, has asymmetrical structures and morphology that would translate the postural and functional loadings to lateral displacement of the patella. Clinically, a person with weakened or unconditioned medial patellar stabilizers may be prone to excessive lateral patellar tracking, PFPS, or even patellar subluxation or dislocation.

1.5. Purpose of the present research

Although various treatments have been prescribed for anterior knee pain, information on the prevention of PFPS is lacking. As measurable predisposing factors of PFPS have been identified, an alteration of the factors may help prevent PFPS; the extent of the alteration that may be generated by weight training is the overall purpose of this thesis. In other words, this thesis explores the physiological and biomechanical changes induced by supervised weight training among previously untrained persons, in terms of strength, morphology and functioning of the passive and active patellar stabilizers. The findings of this thesis could form a basis for the establishment of a PFPS-prevention exercise program and enhance our understanding of the bio-kinesiology of the patellofemoral joint under influence of systematic physical conditioning.

1.6. Organization of this thesis

This thesis consists of three parts that cover seven chapters including the present one. The layout is illustrated on following page (fig 1-9).

Part I

Chapter 1 Introduction

Part II

Validity and Methodological Development

Chapter 2 (Study I) examining the optimal electrode placement for EMG recording of VMO and distal VL.

Chapter 3 (Study II) comparing a new method of determining the EMG onset with an established method.

Chapter 4 (Study III) designing a reliable device for quantifying passive lateral patellar mobility, and investigated the effect of geometrical characteristics of patellofemoral joint on passive patellar mobility.

Chapter 5 (Study IV, V) examining the validity of MRI based measurement for patellar position; and the reliability of the MRI and ultrasound for the cross-sectional area of the active patellar stabilizers.

Main study

Chapter 6 (Study VI) comparing the effect of two different modes of weight training on the active and passive patellar stabilizers.

Part III

Chapter 7 Grand discussion of the present thesis

Figure 1-9: Organization of the present thesis.

Chapter 2

(Study I)

Optimal Positioning of Electrodes for Surface EMG Examination of Vasti Muscles

2.1. Background and literature review

Electromyography (EMG) is the recording of motor unit action potentials or muscle compound action potentials. Surface EMG has been widely used in studies of skeletal muscle activities and quantifying the muscle recruitment and force productions (Basmajian & De Luca 1985a; De Luca 1997; Farina 2006), also the EMG is one of the main outcome measurements in the main study (Chapter 6) of this thesis.

The surface EMG represents an extracellular view of changes in sacrolemma potential related with the propagation of action potentials (Farina 2006). The source of EMG signal is the alpha motor neurons and its motor units. A motor unit is composed of a cell body, its axon and muscle fibers that it innervates. When an action potential is initiated by the motor neuron, it travels down the axons and reaches the motor end plate also known as neuromuscular junction, and then triggers the release of acetylcholine which increases the permeability of sacrolemma (Deschenes et al 1994; Green 2004). The increased permeability of sacrolemma makes the charged ions mobile across the membrane as an electrical current flow occurs.

When a muscle is relaxed, electrical potential across the sacrolemma remains relatively constant and is regarded as negative inside and positive outside due to more negatively charged ions (e.g. hydrogen carbonate) internally and more positively charged ions (e.g. sodium) externally. Because the human body is electrically conductive, the current flow is propagated to adjacent tissues. If a pair of bipolar electrodes is placed over a contracting muscle and preferably parallel to the orientation of muscle fibers, the current researching the electrodes can be detected and electronically processed. A differential amplifier connected to the bipolar electrodes can operate for the differential signal gain and common signal removal, thus the surface EMG amplitude and frequency spectrum can be calculated (Basmajian & De Luca 1985a; De Luca 1997; Farina 2006).

The relative EMG of vastus medialis obliquus (VMO) and vastus lateralis (VL) has been well studied in the past. As the VMO and VL muscles have different morphological, functional, and neuromotor characteristics (Lieb & Perry 1968; Karst & Willett 1995; Powers et al 1996; Crossley et al 2001; Ng 2002), the two muscles work synergistically to provide the dynamic stabilizing force on the patella during knee extension and imbalance between the VMO and VL could disturb the dynamic stability of patella. Clinically, the weakness of the VMO may result in excessive lateral tracking of patella, which has been regarded as a common cause of PFPS (Mariano & Caruso 1979; McConnell 1996; Herrington 1998). Furthermore, some studies reported that people with PFPS had reduced VMO:VL electrical activity ratio (Souza & Gross 1991; Boucher et al 1992, Tang et al 2001) or delayed VMO onset when compared with asymptomatic controls (Voight & Weider 1991; Witvrouw et al 1996; Cowan et al 2001).

Souza & Gross (1991) reported the averaged VMO:VL activity ratio in healthy subjects to be approximately 0.52:1 during isometric knee extension which was significantly different from that of patients with PFPS who demonstrated a ratio of 0.36:1. Boucher et al (1992) reported the ratio to be approximately 2:1 for asymptomatic subjects in static terminal knee extension, whereas the ratio for people with PFPS was 1.5:1. Tang et al (2001) demonstrated that the normalized ratio was approximately 1:1 for normal subjects at the 60° knee flexion position during eccentric 120°/sec isokinetic knee extension while PFPS subjects showed that their ratio was around 0.8:1, a statistical difference was noted between these groups. The above authors stated that patients with PFPS might differ from asymptomatic individuals with regard to VMO: VL activation pattern.

In terms of the temporal parameters of the vasti muscle activation, Voight & Weider (1991) and Witvrouw et al (1996) reported that VL fired significantly earlier than VMO in the knee jerk reflex in patients with PFPS. Their finding was in contrast to Karst & Willett (1995) who reported no difference in VMO-VL contraction timing in the knee jerk reflex in both normal subjects and those with PFPS. Powers et al (1996) also found that the ratio of VMO:VL activity and the VMO-VL relative onset time were not different between patients with PFPS and the normal controls during walking or climbing stairs.

Some researchers attempted to develop specific strengthening exercises for the VMO. Hanten & Schulthies (1990) and Laprade et al (1998) reported that VMO was more active than VL during isometric knee extension with hip adduction and tibial internal rotation. Ng & Man (1996) and Lam & Ng (2001) reported that VMO was significantly more active than VL in static knee extension and squat with internal hip rotation. However, Reynolds et al (1983) and Mirzabeigi et al (1999) did not support the concept of selective recruitment of VMO over VL in physical training.

The discrepancies of the above findings could be due to the EMG instruments and electrode placements not being comparable among these studies. Boucher et al (1992) and Tang et al (2001) placed an electrode over the motor point of VL at the mid thigh level which was approximately 15 cm above the upper border of patella. Gilleard et al (1998) and Ng (2002) placed the VL electrode at 10 cm above the patella. Souza & Gross (1991), Cram & Kasman (1998) and Willis et al (2005) reported that the electrode for the VL muscle was at 5 cm or 6 cm above the patella. Conversely, the electrode placement for VMO was more consistent among previous studies as most reported the position to be 3 or 4 cm above the supero-medial corner of patella (Gilleard et al 1998; Ng 2002) (fig 2-1).

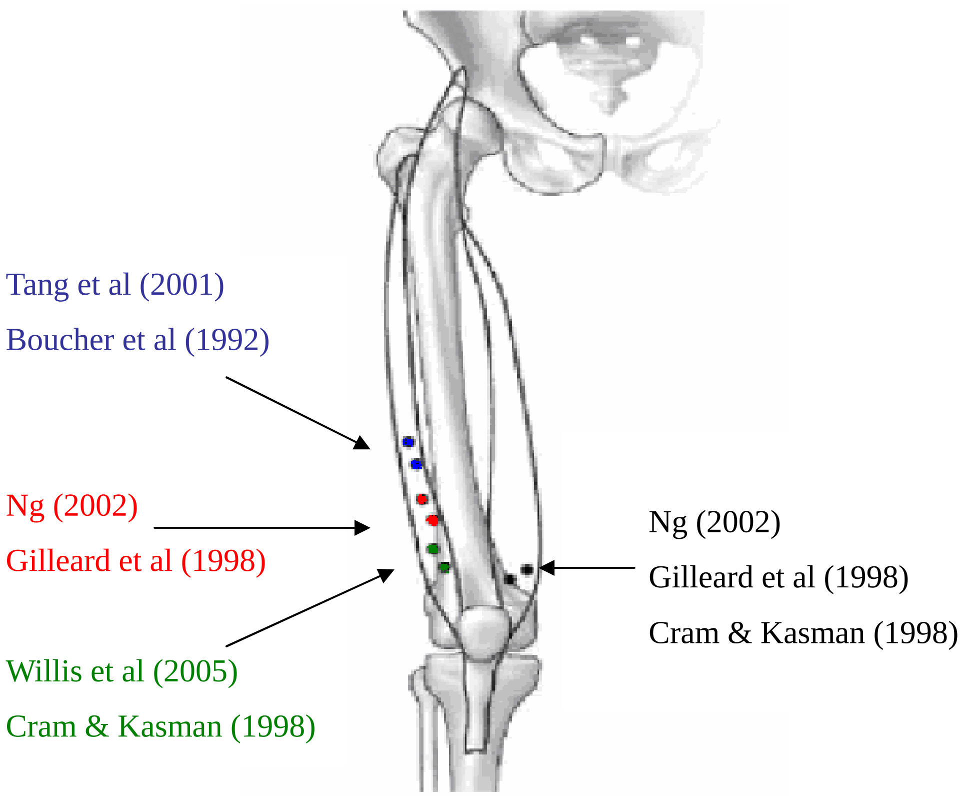


Figure 2-1: Different electrode placement on VMO and VL were used previously by various researchers.

2.2. Objectives

In light of the diverse positioning of the VL electrode and the lack of information on the effects of surface electrode positioning on the EMG recording, there is a need to examine if electrode positioning would affect the EMG measurements. This has vital clinical implications because imbalanced VMO:VL EMG activity or VMO-VL onset time is regarded as contributing factors of PFPS, the effect of electrode positioning must be considered and controlled in order to use the EMG data for clinical assessments. Therefore, a study was conducted to investigate the effects of different surface electrode placements on the VMO:VL activity ratio and VMO-VL relative onset timing. This study was published in Physical Therapy in Sports in 2006 (Appendix 7).

2.3. Hypothesis

Different EMG electrode placements on the VMO and VL would lead to different EMG recordings in terms of VMO:VL ratio and VMO-VL onset time.

2.4. Methods

<u>Subjects</u>

Eight able-bodied non-athletic males aged between 24 and 35 years (mean 28.1 years) were recruited. Subjects with a history of lower limb operations or injuries that required treatments in the past 6 months were excluded. The study was reviewed and approved by the Human Ethics Sub-committee of the Hong Kong Polytechnic University prior to data collection (Appendix 1). All subjects were asked to provide informed written consent (Appendix 2) and refrain from vigorous physical activities on the day before the study, so as not to induce fatigue to the muscles.

List of apparatus

1) Flexcomp computerized data acquisition system (Thought Technology, Montreal, Quebec, Canada) (figs 2-2, 2-3).
2) Single-differential EMG electrode with 500 gain, common mode rejection ratio > 130 dB at 50 Hz, input impedance = 10 Giga-ohm (Myoscan, Thought Technology, Montreal, Quebec, Canada) (figs 2-3, 2-4).
3) Disposable silver/silver chloride surface adhesive electrode with a 10 mm circular contact area and an inter-electrode distance of 20 mm (Multi Bio Sensors, El Paso, TX, USA) (fig 2-4).
4) Skin abrasive gel (NuPrep, DO Weaver & Co., Aurora, CO, USA).
5) Conductive gel (Ultra/Phone, Pharmaceutical Innovations, Newark, NJ, USA)
6) Skin impedance meter (1089MKIII, UFI, Morro Bay, CA, USA) (fig 2-5).
7) Isometric/isokinetic dynamometer (Cybex Norm, Henley Healthcare, Nauppauge, NY, USA) (fig 2-6)

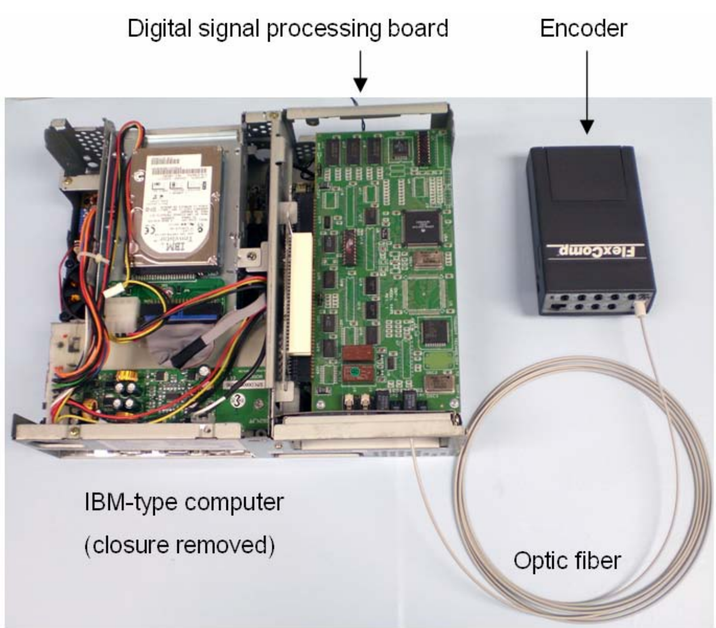


Figure 2-2: Flexcomp computerized data acquisition system.

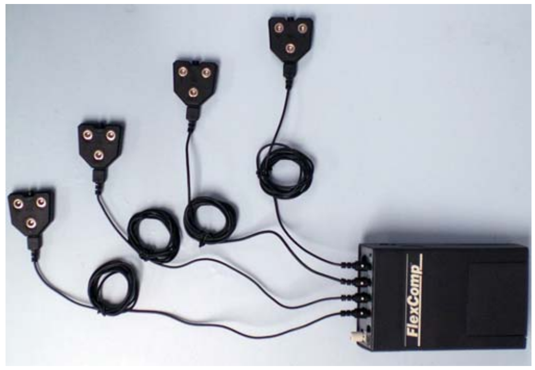


Figure 2-3: Flexcomp encoder connected with EMG electrodes.

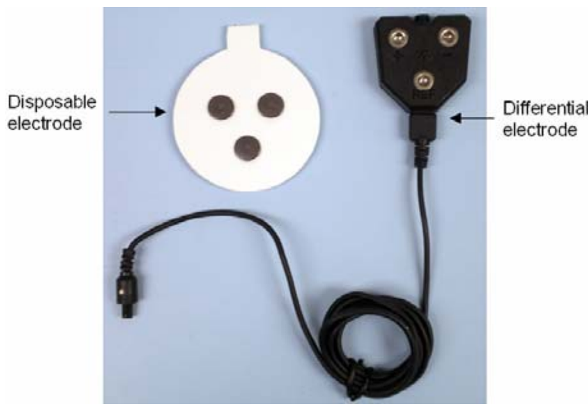


Figure 2-4: Disposable electrode and surface electrode with built-in differential amplifier.

Figure 2-5: Skin impedance meter.

Figure 2-6: Isometric/isokinetic dynamometer.

EMG recordings

Skin on the distal thigh of the dominant leg was shaved and cleansed with skin abrasive gel (NuPrep, DO Weaver & Co., Aurora, CO, USA) and methylated spirit. The skin impedance was checked with a skin impedance meter (1089MKIII, UFI, Morro Bay, CA, USA) (fig 2-5) and a value of less than 50 kΩ was deemed acceptable (Hewson et al 2003). After the skin preparation, four single-differential EMG electrodes (Myoscan, Thought Technology, Montreal, Quebec, Canada) (fig 2-3) were placed over the VMO and VL with disposable snap electrodes (Multi Bio Sensors, El Paso, TX, USA) (fig 2-4). A thin layer of conductive gel (Ultra/Phone, Pharmaceutical Innovations, Newark, USA) was put on the electrode surfaces. The electrode for VMO was placed at a point 4 cm above the supero-medial corner of patella, whereas three electrodes were placed over VL at 5 cm (electrode 1), 10 cm (electrode 2), and 15 cm (electrode 3) above the supero-lateral corner of patella (fig 2.7). The orientations of the VMO and VL electrodes were 55° and 15° to the longitudinal axis of femur, respectively (Lieb & Perry 1968).

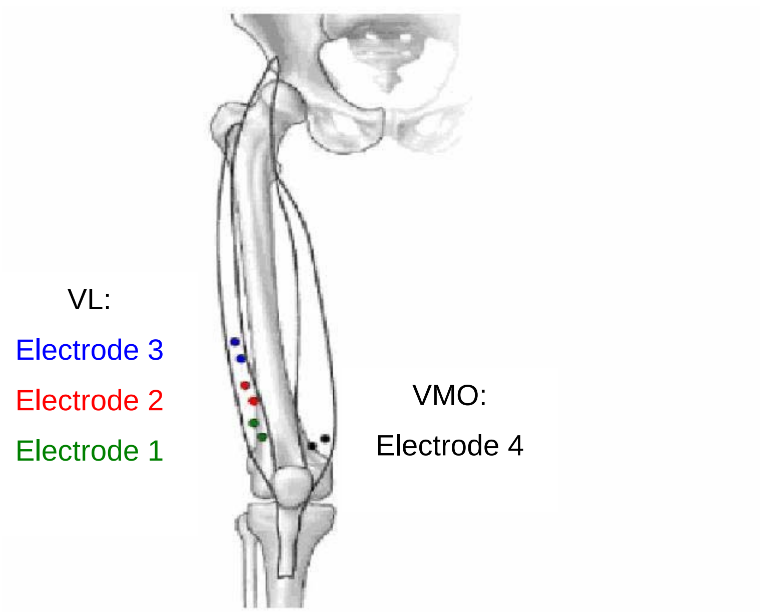


Figure 2-7: Positioning of the EMG electrodes for VL and VMO of the right leg.

The electrodes were connected to a battery powered encoder (SA9404P, Thought Technology, Montreal, Quebec, Canada) (figs 2-2, 2-3) with a sampling rate of 1984 Hz. The encoder was fed to a digital signal processing board (DSP, version 2, Thought Technology, Montreal, Quebec, Canada) (fig 2-2) installed on a personal computer (Celeron 400MHz, Windows 98). A data-acquisition software (Ergonomic Suite V1.52, Thought Technology, Montreal, Quebec, Canada) was used to capture the raw EMG with bandwidth of 20-500Hz. The raw data were later analyzed off-line with Excel software (Office 2000, Microsoft, Redmond, WA, USA).

Isometric knee extension

An isometric/isokinetic dynamometer (Cybex Norm, Henley Healthcare, Nauppauge, NY, USA) was used to measure the isometric maximal voluntary contraction (MVC) on the dominant leg of each subject with the knee at 45° and hip at 85° of flexion (fig 2-8). The dominant leg was defined as the one the subjects used to kick a football. Subjects were requested to perform 5 repetitions of submaximal knee extension to get acquainted with the isometric testing prior to the MVC test that lasted for 4 seconds. The test was repeated 3 times with a 2-minute rest between each session.

The highest force developed in the 3 tests was regarded as the MVC. Afterwards, the subjects performed 5 seconds of isometric knee extension at 50% and 75% of their MVC with real-time visual signals from the dynamometer screen to guide them on the force output. These submaximal tests were implemented once because the EMG amplitude measurement with submaximal muscle contraction was found to be highly repeatable in short-term (Kollmitzer et al 1999). During the isometric testing, EMG activities of VMO and VL were recorded with the above data acquisition system and computer.

Knee perturbation test

After the knee extension tests, the subjects proceeded to a knee perturbation test with single leg standing so as to induce a stretch reflex in the quadriceps muscles (Ng 2005). Subjects stood on their dominant leg with the knee extended and the hands lightly touching the wall in front for balance. A medicine ball of 3.5 kg was used to provide perturbation to the knee joint. The ball was suspended with a sling from a hook at the ceiling. As the length of the sling was adjustable, the medicine ball was adjusted to be at the level of the knee joint of the subjects (fig 2-9). The examiner pulled the ball back until the sling made an angle of 45^o to the vertical behind the subject and upon release of the ball, it would exhibit a pendulum swing and hit on the back of the subject's knee to produce a perturbation. The perturbation test was repeated 3 times with 1 minute in-between while the EMG of VMO and VL were recorded.

Figure 2-8: Isometric knee extension.

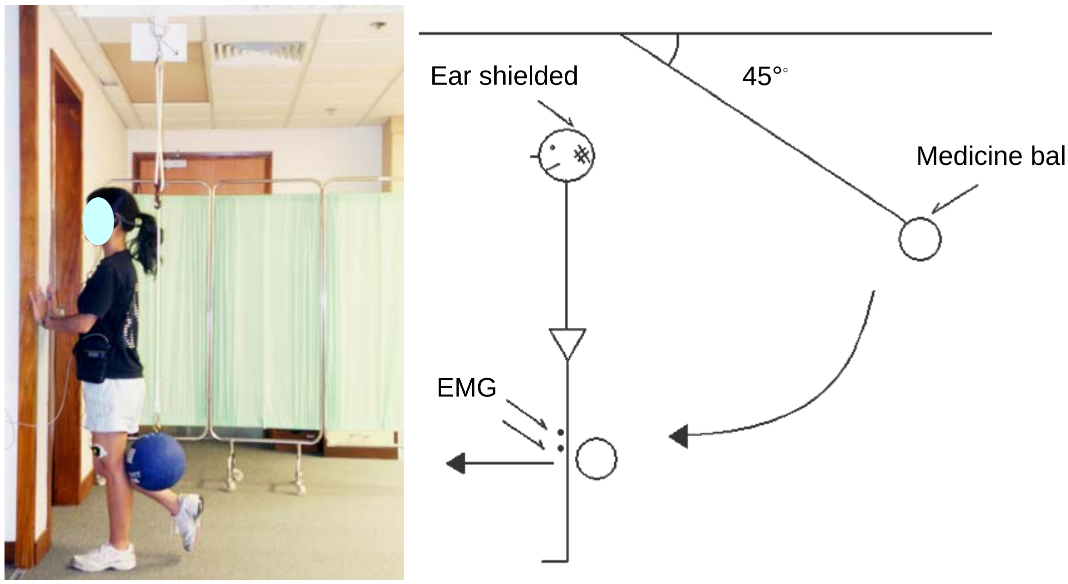


Figure 2-9: Knee perturbation test.

<u>Data analysis</u>

For the isometric knee extension, the EMG signal was transformed to root-mean-square (RMS) in time constant of 1000 ms. The initial 2 seconds of EMG data were excluded so as to avoid the fluctuation of EMG signal. Then each subject's highest RMS value of VMO and VL at 50% and 75% MVC were normalized against the respective muscle's 100% MVC (Soderberg & Knutson 2000). Since there were 3 VL electrodes and 1 VMO electrode, 3 normalized VMO:VL ratios were obtained for each MVC testing condition.

For the knee perturbation test, the EMG data were full-wave rectified and the means and standard deviations of the 100-ms resting EMG signal prior to perturbation were calculated. The EMG onset was determined as a signal that had 2 standard deviations above the resting mean for 10 ms or more (McKinley & Pedotti 1992; De Luca 1997). Relative VMO-VL onset time was calculated by as the onset of VL minis that of the VMO. Since there were 3 electrodes on the VL, 3 VMO-VL relative EMG onset times were obtained. The mean EMG onset time of the 3 perturbation trials was calculated for each subject in order to minimize random error, and intraclass correlation coefficient (ICC (3,1), absolute agreement, CI 95%) was calculated to determine its within-day reliability.

Repeated measures ANOVA ($p = 0.05$) with Bonferroni adjustment for within-subject comparisons (SPSS 11.5.1, SPSS Inc., Chicago, IL, USA) was used to analyze the normalized VMO:VL activity ratio and the relative VMO-VL onset timing difference among the 3 pairs of the VMO and VL electrodes.

2.5. Results

The ICC (3,1) values of the EMG onset timing difference of VMO-VL in the knee perturbation measurement were 0.86, 0.87, and 0.87 for the electrode pairs of 1 and 4, 2 and 4, and 3 and 4 respectively. These ICC values indicated that the EMG recordings had good to very good reliability (Portney & Watkins 2000). The means and standard deviations of the VMO-VL onset time and VMO:VL EMG activities are listed in table 2-1.

Table 2-1: Dependent variables in isometric knee extension and knee perturbation. The means (standard deviations) of EMG onset timing during perturbation and EMG activity ratios of VMO:VL during 50% and 75% of MVC testing

Electrode pair	Normalized VMO:VL ratio		VMO-VL onset time (ms)
	50% MVC	75% MVC	
1 & 4	1.19 (0.04)	1.24 (0.07)	6.70 (3.35)
2 & 4	0.97 (0.08)	1.05 (0.08)	8.91 (4.96)
3 & 4	0.90 (0.09)	0.98 (0.08)	12.28 (7.10)

Results of repeated measures ANOVA revealed significant differences in both the VMO:VL ratios and VMO-VL onset time (table 2-2). Post-hoc analyses revealed that the recording with electrode pair 1:4 was significantly different from the electrode pairs of 2:4 and 3:4 for the onset timing. For the VMO:VL activity ratio, electrode pair 1:4 had higher ratios than 2:4 and 3:4 at 50% activation level, whereas at 75% of activation level, both 1:4 and 2:4 had higher ratios than 3:4 (table 2-2).

Table 2-2: Results of repeated measures ANOVA with Bonferroni adjustment of EMG data. As EMG activity ratios of VMO:VL during 50% and 75% of MVC isometric knee extension and EMG onset timing during perturbation

Measurements	ANOVA	Pair-wise comparison
VMO:VL ratio in 50% MVC	df =1.028, F = 21.157, p = 0.002	Electrode pair of 1:4 > 2:4 (p=0.018) and 3:4 (p=0.004); 2:4 > 3:4 (p<0.0001)
VMO:VL ratio in 75% MVC	df =1.044, F = 10.121, p = 0.014	Electrode pair of 1:4 and 2:4 > 3:4 (p=0.018, 0.004)
VMO-VL onset difference in knee perturbation	df = 2, F = 10.555, p = 0.002	Electrode pair of 1:4 < 2:4 (p=0.03) and 3:4 (p=0.004)

2.6. Discussion

This study aims to examine the effect of surface electrode positioning on the EMG recordings of VMO and VL with respect to the onset timing during quadriceps stretch reflex and EMG voltage during voluntary isometric contraction. Results revealed that different positioning of the VL electrodes would significantly affect the EMG onset time and electrical signal strength during the same activity. The present findings may explain the discrepancies in the previous studies that reported inconsistent VMO:VL activity ratio and VMO-VL onset timing (Souza & Gross 1991; Voight & Weider 1991; Boucher et al 1992; Karst & Willett 1995; Witvrouw et al 1996; Laprade et al 1998; Mirzabeigi et al 1999; Tang et al 2001). The present results suggested that during submaximal muscle contraction, the EMG signals obtained from the electrodes 2 and 3 which were closer to the VL innervation zone at mid thigh level (Warfel 1985) had relatively lower electrical strength and delayed EMG onset time. These findings are parallel to the reports of Souza & Gross (1991) and Boucher et al (1992). With the former study positioning the electrodes on mid VL and reported the VMO:VL ratio to be 2:1, while the latter study positioned the electrode on the distal VL and reported the ratio to be 0.65:1 for healthy subjects during terminal isometric knee extension.

The EMG electrodes used in this study contained a differential amplifier with two recording and one reference terminals. The differential amplifier would magnify the non-identical signals between the two input terminals and reject any signal which is common to both (Peek 1987). Theoretically, no EMG signal can be detected if the two recording terminals are placed equally on both sides of the motor end-plate of a muscle along its fiber direction. This is because when a muscle action potential is generated at the motor end-plate and propagates to both ends of the muscle, the signals will be registered as identical by both input electrode terminals thus cancelled by the differential amplifier (Campbell 1999). However, this never happens with *in vivo* measurement, because the electrodes detect muscle action potentials generated by different motor end-plates instead of from a single motor end-plate. These signals reach the input terminals at slightly different time intervals, thus complete cancellation does not occur (Basmajian & De Luca 1985a).

Electrodes located midway between the motor points and tendinous insertion would result in relatively larger signal amplitude since the action potentials would reach the input terminals with very different phasic properties (Kleine et al 2001). It has been reported that the EMG amplitudes of upper trapezius and tibialis anterior muscles would be lower when the electrodes were placed near the innervation zones during both voluntary contraction and electrical stimulation (Jensen et al 1993; Merletti et al 1993). This may explain our present findings that electrode 3 located in the middle of VL had the lowest signal strength because it is where most of the motor endplates are found and they cluster around this electrode position (Warfel 1985).

For the relative VMO-VL onset timing, results revealed that the electrode pairs 3 & 4 and 2 & 4 had a significantly greater difference than the pair 1 & 4. This is in agreement with the finding on the EMG amplitude because the determination of the onset was dependent on the difference between the voltage of resting muscle electrical signal and that of the activated muscle action potential. With a relatively lower action potential strength detected in positions 3 and 2, the rate of voltage build-up in this position would also be slower, thus the temporal activation point was registered later.

These findings indicate that the EMG activity ratio and onset timing for the VMO and VL are electrode position dependent. Studies using unique electrode locations over the vasti muscles could result in incompatible outcomes. However, the finding is restricted to surface electrodes; it is not known if the same finding can be applied to fine wire EMG as the wire electrode intramuscularly detects one or few individual motor unit action potentials rather than multiple motor unit action potentials monitored at top of skin in the surface EMG (Basmajian & De Luca 1985a).

One of the limitations in the present study is the lack of external validity because only asymptomatic subjects were tested, thus the findings may not be directly applicable to people with PFPS. In addition, all subjects' knee joints were either positioned at 45° flexion or full extension; it is natural that if the knee joint flexes by more than 45°, the vasti muscles will be stretched further, thus the innervation zone of VMO and VL may shift inferiorly and the EMG characteristics may change accordingly.

In the present study, the variability of subjects' height was less than 8 cm. Thus the electrodes being positioned on a few spots on the quadriceps with standardized distance from the patella may not be a confounding factor for the EMG recording. However, for clinical measurements, the subjects' height could vary significantly; the use of a percentage of limb length may ensure the standardization of the electrode placements for individuals. This may also enhance between-subject and between-study comparisons.

2.7. Conclusion

The positions of surface electrodes would affect the EMG recording of VMO and VL. Physical therapists or researchers should be aware that a difference of a few centimeters in the position of the VL electrode could affect the VMO:VL EMG activity ratio and VMO-VL onset timing.

2.8. Relevance to the main study

In the main study of this thesis (Chapter 6) that investigated the effect of weight training on the EMG of vasti muscles, the electrode for VMO was placed at 4 cm above the supero-medial corner of patella, whereas the electrodes for VL was placed at 5 cm above the supero-lateral corner of patella; the electrodes for VMO and VL are placed at the similar level so that the distal muscle fibers of VM and VL that work synergistically and directly pulling the patella can be monitored. The orientations of the VMO and VL electrodes were 55° and 15° to the shaft of femur. Thus, the electrical activity of the VMO and VL adjacent to the patellofemoral joint could be compared consistently throughout the thesis.

Chapter 3

(Study II)

Development of a New Method for EMG Onset Detection

3.1. Background and literature review

Detection for the onset of skeletal muscle activities is often required in the treatment of musculoskeletal disorders, muscle training and gait analysis (Winter 1984). For determining the onset of muscle contraction, most clinicians rely on their visual inspection, palpation and experience (Kendall et al 1993). However, these observations are rather subjective and their reliability is often doubtful (Di Fabio 1987). With the need of improving the objectivity and reliability of muscle onset determination, researchers have used EMG technique to study the skeletal muscle activities. In order to accurately quantify the muscle onset time, various quantitative and computerized methods of analyzing the EMG signals have been developed in the past three decades (Di Fabio 1987; Hodges & Bui 1996; De Luca 1997; Abbink et al 1998).

The most popular method for determining the EMG onset was described by Hodges & Bui (1996). They processed the EMG signals with 50Hz low-pass filter and rectification; the time of muscle onset was set as the instance at which the EMG signal was at three standard deviations above the mean (mean + 3 s.d.) of the baseline signal for at least 25 ms. This method has since been widely used by various researchers for studying different muscles under various testing modes (Cowan et al 2001, 2002a,b,c, 2003; Lam & Ng 2001; Ng & Cheng 2002; Hinman et al 2002; Crossley et al 2004; Ng & Chan 2004; Ng 2005). For determining the onset of VMO (vastus medialis obliquus) and VL (vastus lateralis), the method developed by Hodges and Bui (1996) is also most widely used (table 3-1).

Despite its popularity, analysis of the EMG onset time with that method requires the original EMG signals to be exported to a computer for mathematical processing, which may not be convenient for clinicians to perform in the clinical settings. Furthermore, many EMG systems used by clinicians would only display the EMG waveforms, thus mathematical processing of the data could not be done online or even offline. Therefore, if a quantitative, reliable and convenient method for determining the muscle onset time can be established, it will benefit EMG users in both clinical and research settings.

Vaes et al (2001, 2002) were the first group of researchers that reported the EMG onset time of the peroneal muscle by using the peak-to-peak amplitude of the surface EMG signals at double of that of the baseline values (DP-P). This method does not involve sophisticated EMG signal processing techniques, thus clinicians can trace the EMG output signals with existing software or use an electrocardiographic caliper to locate the onset point on the printed waveform (fig 3-1). Therefore, it can be economically implemented in almost any EMG system that does not have statistical or data exporting functions.

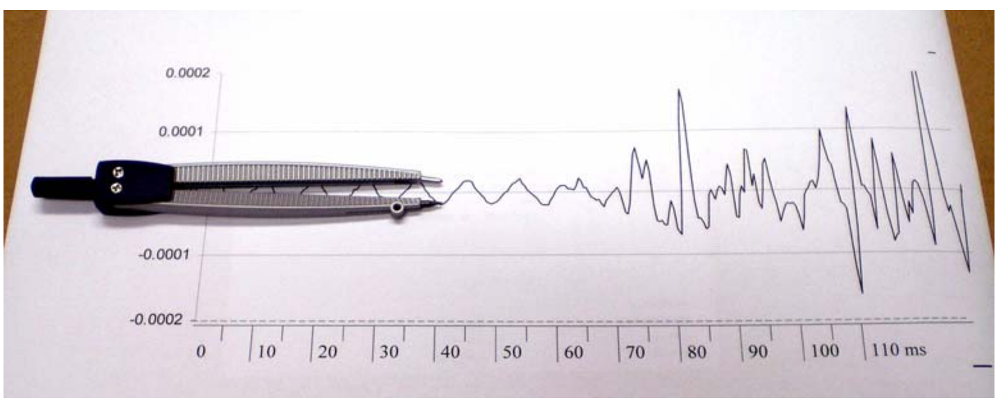


Figure 3-1: Electrocardiographic caliper can be used to locate EMG onset point as at double of the baseline amplitude.

Table 3-1: Summary of EMG onset determination method for VMO and VL (N/A = not applicable)

Author(s)	Subjects		VMO/VL onset induced by	VMO and VL onset threshold
	PFPS	Healthy		
Voight & Weider 1991	16	41	Knee jerk reflex	Visual inspection
Karst & Willett 1995	15	12	Knee jerk reflex, active knee extension	1 standard deviation (SD) above mean of rectified baseline signal
Witvrouw et al 1996	19	80	Knee jerk reflex	Visual inspection
Sheehy et al 1998	13	15	Ascent and decent stair-stepping	Peak amplitude of RMS with time constant of 50ms
Cowan et al 2001	33	33	Ascent and decent stair-stepping	3 SD above mean of rectified baseline signal + 50Hz low-passed
Cowan et al 2002a	65	N/A	Ascent and decent stair-stepping	3 SD above mean of rectified baseline signal + 50Hz low-passed
Hinman et al 2002	41	33	Ascent and decent stair-stepping	3 SD above mean of rectified baseline signal + 50Hz low-passed
Cowan et al 2002b	10	12	Stair-stepping with and without patellar taping	3 SD above mean of rectified baseline signal + 50Hz low-passed
Cowan et al 2002c	37	37	Toes tipping and heel rising in standing	3 SD above mean of rectified baseline signal + 50Hz low-passed
Brindle et al 2003	22	12	Ascent and decent stair-stepping	5 SD above mean of rectified baseline signal
Cowan et al 2003	40	N/A	Toes tipping and heel rising in standing	3 SD above mean of rectified baseline signal + 50Hz low-passed
Crossley et al 2004	48	18	Ascent and decent stair-stepping	3 SD above mean of rectified baseline signal + 50Hz low-passed
Ng 2005	N/A	29	Knee perturbation with and without patellar taping	3 SD above mean of rectified baseline signal
Crossley et al 2005	40	N/A	Ascent and decent stair-stepping	3 SD above mean of rectified baseline signal + 50Hz low-passed
Hinman et al 2005	N/A	66	Ascent and decent stair-stepping	3 SD above mean of rectified baseline signal + 50Hz low-passed

3.2. Objectives

Since the reliability of the double peak-to-peak (DP-P) method and the compatibility of the results of this method with other established methods such as the mean + 3 s.d. have not been established, particularly for the patellar active stabilizers VMO and VL, this study aimed to:

1) examine the reliability of identification of EMG onset using the DP-P method, and
2) compare the results of this method with that of the mean + 3 s.d. method for the muscles of the knee in able-bodied subjects.

If the DP-P method of muscle onset detection was found to be reliable and the results were comparable with the other method, it could be applied for studying the VMO and VL activation of subjects with or without different pathologies in both clinical and research settings. This study was published in Electromyography and Clinical Neurophysiology in 2005 (Appendix 8).

3.3. Hypothesis

The new DP-P EMG onset detection is repeatable and comparable to the traditional method of mean + 3 s.d.

3.4. Methods

<u>Subjects</u>

Eleven subjects (five females and six males) with the mean age of 28.4 (+/- 4.8, ranging 22-35) volunteered to participate in this study. All subjects were healthy with no history of lower limb surgery and physical injuries that required medical treatments for at least 6 months prior to this study. The study was reviewed and approved by the Human Ethics Sub-committee of the Hong Kong Polytechnic University prior to data collection (Appendix 3). All subjects were asked to provide written informed consent (Appendix 4) and abstain from vigorous physical activities on the day before the study.

<u>List of apparatus</u>

1) Apparatus used in this study were the same as those used in Chapter 2 (Study I) (fig 3-2).

2) Voltage isolator (T9405, Thought Technology, Montreal, Quebec, Canada) (fig 3-2).

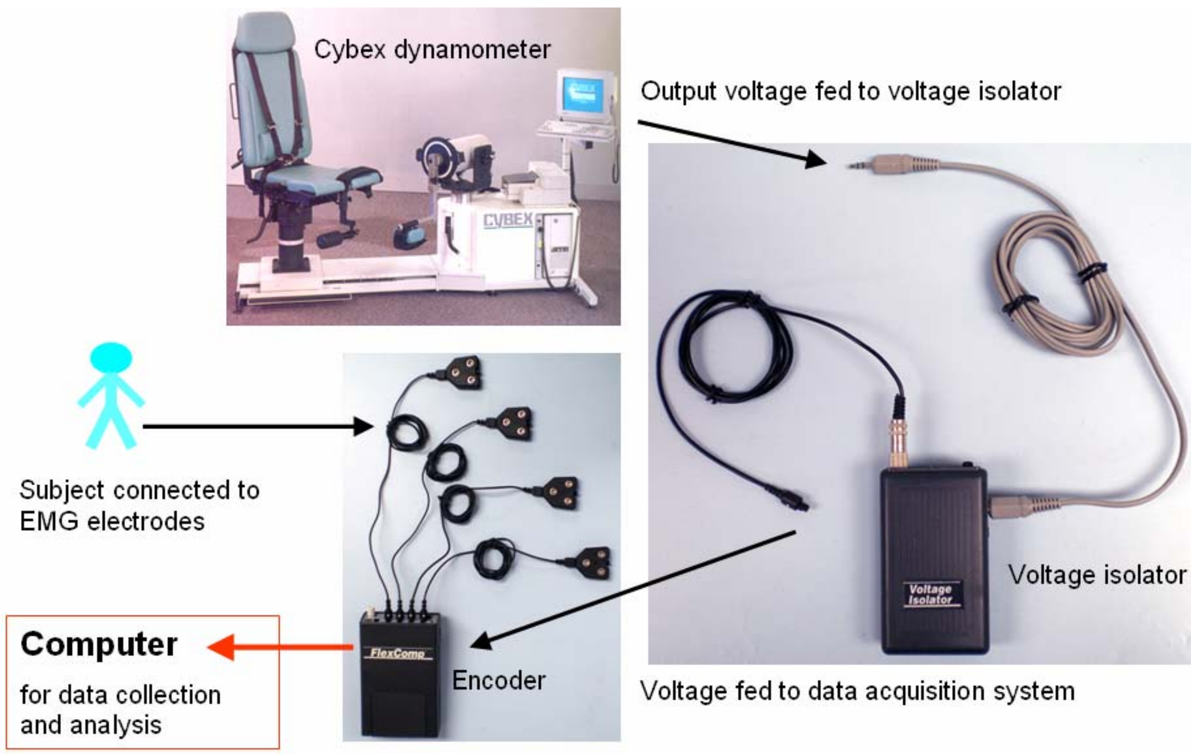


Figure 3-2: EMG and isometric force data acquisition system. The voltage isolator acted as a surge protector for subjects who were indirectly connected with the AC (220V) powered Cybex dynamometer, in case of the dynamometer malfunctions, its leaking AC voltage would not reach the subject.

Isometric torque and EMG recordings

A dynamometer (Cybex Norm, Henley Healthcare, Nauppauge, NY, USA) was used to measure the isometric knee extension strength of the subjects' left leg at 45° of knee flexion and 85° of hip flexion. The knee extension torque values were converted to DC voltages (1 V = 80 Nm) and transmitted to a battery powered encoder (SA9404P, Thought Technology, Montreal, Quebec, Canada) at a sampling rate of 124 Hz via a voltage isolator (T9405, Thought Technology, Montreal, Quebec, Canada) (fig 3-2). The encoder was also connected to a digital signal processing board (DSP, version 2, Thought Technology, Montreal, Quebec, Canada) installed in a personal computer (Celeron 400MHz, Window 98). Data acquisition and analysis software (Ergonomic Suite V1.52, Thought Technology, Montreal, Quebec, Canada) was used to provide real-time visual feedback of the torque output to subjects and for off-line data analyses.

Two single-differential surface electrodes (SA9503M, Thought technology, Montreal, Quebec, Canada) were connected to the encoder for EMG data capturing. The sampling rate for the EMG data was 1984 Hz.

This study took place in a laboratory and only one subject was tested at a time. After all the procedures of the EMG measurement were explained and demonstrated to the subject, the skin over the VMO and VL of the subject's dominant leg was prepared by shaving and cleansing with skin preparation gel (NuPrep, DO Weaver & Co., Aurora, CO, USA) and alcohol.

The electrode placement was based on the findings reported in Chapter two. The electrodes for VMO were positioned at 4 cm superior to the superomedial patellar border and aligned at 55^{o} to the longitudinal femoral axis, whereas the electrodes for VL recording were positioned 5 cm superior and aligned at 15^{o} to the femur. The electrodes were secured on VMO and VL by disposable circular snap electrodes (Multi Bio Sensors, El Paso, TX, USA) with a standard 2 cm spacing of silver/silver chloride electrodes that have pickup areas of 1 cm in diameter. A thin layer of conductive gel (Ultra/Phone, Pharmaceutical Innovations, Newark, USA) was put on the pickup surfaces.

After the above preparations, each subject performed 3 maximal isometric knee extensions. The highest isometric extensor torque of the 3 contractions was registered as the maximal voluntary contraction (MVC) of the subject. After registering the MVC, the subject would be given 2 minutes of rest and then the subject was requested to perform the following actions in random order:

1) 100% MVC knee extension.

2) 75% MVC knee extension.

3) 50% MVC knee extension.

Visual feedback was provided to the subjects to help them develop the required levels of MVC. A value of +/-10% around the target level of MCV was regarded as acceptable for these trials. The EMG of the vasti muscles were repeatedly recorded in three occasions. The first two recordings were conducted with a 30-minute interval without removing the electrodes whereas the third recording was at 7 days later (fig 3-3).

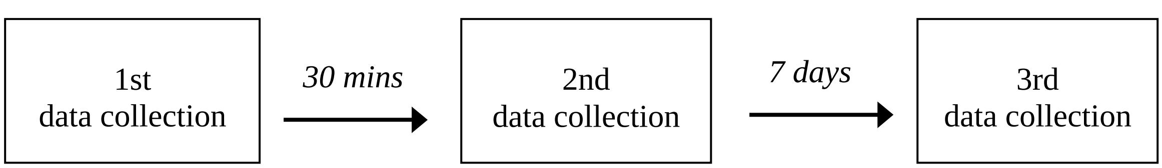


Figure 3-3: Logistics of data collections each included the EMG at three levels of MVC.

Data analysis

In the isomeric knee extension trials, three synchronized signals were acquired as two AC voltages for EMG of the VMO and VL, and one DC voltage for the isomeric knee extension torque. Since the EMG onset of the vasti occurred prior to the onset of the isometric torque known as the electromechanical delay (De Luca 1997; Kubo et al 2001), the onset of the isometric torque was used to act as a reference for calculating the EMG onset of the vasti muscles. Thus, the sudden increase in the DC voltage was regarded as the time of the torque onset which was used to minus the time of the EMG onset of the VMO and VL respectively, and would result in the absolute EMG onsets that could be compared for its test-retest reliability. The onsets were shown in unit of milliseconds (ms).

EMG onset determined by DP-P: The raw EMG data was processed with a filter of 20-500Hz bandwidth, and then an EMG signal of 200-ms in duration that preceded the isometric torque onset was manually selected as the baseline reference. The onset of muscle activation as determined by the DP-P method was regarded as the point at which the EMG peak-to-peak intensity doubled that of the highest peak-to-peak waveform in the baseline reference (Vaes et al 2001, 2002; Weerd at al 2004).

EMG onset determined by mean + 3 s.d.: The raw EMG signal was low-pass filtered with 50 Hz and full-wave rectified. The processed signal was saved at CSV (comma separated values) file and then fed to the Excel software (Office 2000, Microsoft, Redmond, WA, USA) for the mean + 3 s.d. analysis. The onset point was determined as the instance at which the processed signal was higher than 3 standard deviations over the mean value of the reference signal and such signal lasted for 25ms or longer (Hodges & Bui 1996, Cowan et al 2002a,b,c, 2003; Hinman et al 2002, 2005; Crossley et al 2005; Ng 2005).

Test-retest reliability of DP-P: Intra-class Correlation Coefficients (ICC (3,1), absolute agreement, CI 95%) was used to establish the test-retest reliability of the EMG onset time of the vasti muscles being determined by the DP-P.

Relationship between DP-P and mean + 3 s.d.: Linear regression was used to calculate the r^2.

Comparability between DP-P and mean + 3 s.d.: Comparison of the DP-P to the mean + 3 s.d. was done using paired t-test with alpha level set at 0.05.

3.5. Results

The ICC values for the EMG onset time of the vasti muscles being determined by the DP-P is shown in table 3-2. A trend of increasing test-retest reliability with decreasing levels of MVC was noted.

Table 3-2: The ICC values indicating the reliability of the DP-P method.

Interval	100% MVC		75% MVC		50% MVC	
	VMO	VL	VMO	VL	VMO	VL
30-min	0.66	0.64	0.76	0.67	0.86	0.75
7-day	0.66	0.63	0.66	0.67	0.80	0.81

The r^2 values were listed in table 3-3 that showed the DP-P to be fairly predictive to the mean + s.d. However, a significant difference was found between the results of the two methods for both VL and VMO ($p<0.0001$). In the cell point chart that showed the means (standard deviation) of the EMG onset time of VMO and VL (fig 3-4), it can be seen that there was a consistent delay in the DP-P method as compared to that of the mean + 3 s.d. method for the same muscle action.

Table 3-3: r^2 values representing relationships between DP-P and mean + 3 s.d.

	VMO	VL
50% MVC	0.70	0.83
75% MVC	0.57	0.61
100% MVC	0.79	0.65

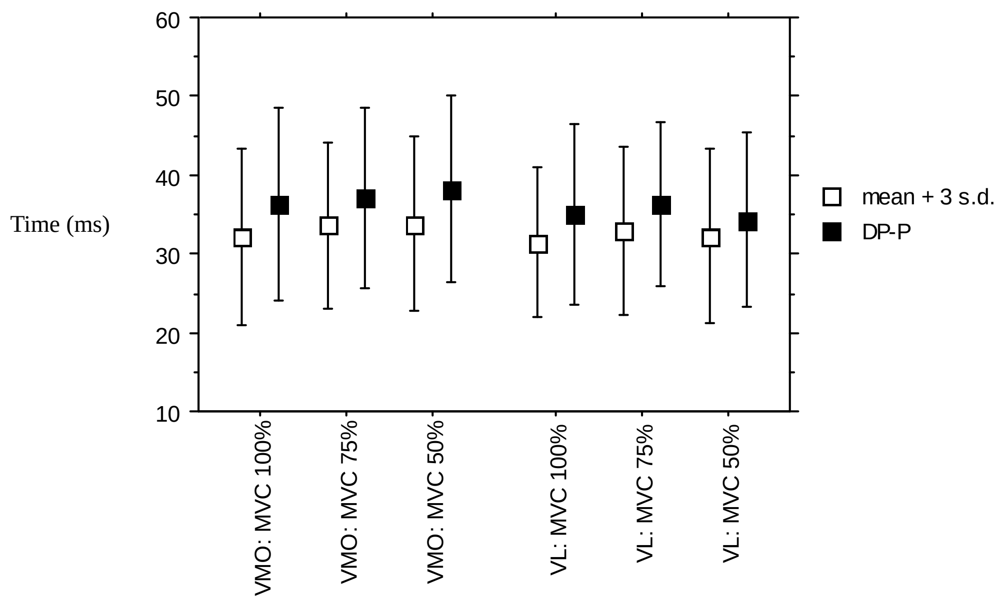


Figure 3-4: Comparison of EMG onset time of contraction of VMO and VL determined by the mean + 3 s.d. and the DP-P methods.

3.6. Discussion

This study aims to examine the test-retest reliability of the DP-P method for EMG onset identification and to compare the results of DP-P method with the mean + 3 s.d. method. The findings reveal that the DP-P method has moderate to good reliability with

ICC (3,1) ranging from 0.63-0.86 (Portney & Watkins 2000). Since EMG measurement is susceptible to surrounding electromagnetic interference that can substantially increase the EMG offset, statistics such as the Pearson's correlation is considered not a suitable index of reliability when there is a substantial difference between two EMG measurements among the subjects and their ranks remain the same in the two measurements. Therefore ICC was used as it can detect both changes in the rank among the subjects tested and the consistency of the measurements across occasions (Arnold 1997).

The method of DP-P revealed that the EMG onset detection was more repeatable in the 50% MVC isometric knee extension than higher levels of MVC. This finding is in line with the report of Kollmitzer et al (1999) in which recordings of root-mean-square EMG amplitude of isometric knee extension taken at 50% MVC demonstrated better repeatability than 100% MVC measurements. The good reliability of this method with submaximal muscle testing is clinically meaningful, because most of the therapeutic exercises or functional assessments of the vasti muscles are done at submaximal levels, and the EMG recordings at these submaximal contractions could be compared periodically.

More importantly, the present results revealed very good between-day reliability of the DP-P method with ICC for between-day comparisons >0.8 at 50% MVC. This finding implies that the effects of exercise training to facilitate muscle onset may be assessed reliably with the DP-P method, which has important clinical application value.

The onset times of VMO and VL have been widely explored with the mean + 3 s.d. method in previous studies (Cowan et al 2001, 2002a,b,c, 2003; Hinman et al 2002, 2005; Crossley et al 2004, 2005; Ng 2005), thus the validity of comparing the DP-P to this well established method is justified. Furthermore, accurate measurements of the onset time of

VMO and VL have vital clinical implications for the management of patients with patellofemoral pain syndrome, because the treatment for these patients often targets to facilitate VMO contraction. Developing a user-friendly detection method for clinicians will facilitate the clinical management of this condition.

However, a point of caution is that the findings on the vasti muscles may not be generalized to other muscles, because surface EMG recordings could be affected by factors such as thickness of subcutaneous tissue, electrode placement, cross talk of adjacent muscles, muscular length and fatigue. In particular, muscles of the limbs work against inertia whereas muscle in other parts of the body such as the chest wall work against elasticity and their EMG/force relationships had been reported to be different (Ng & Stokes 1992). Applications of the findings from lower limb muscles to other muscles may need to be further validated.

When comparing the results of the DP-P with the mean + 3 s.d. method, the DP-P method showed a constant delay of about 3 ms in the onset time of both VMO and VL (fig 3-4). Since the delay is consistent in both muscles and across all conditions of testing, this should not affect the comparability of the relative EMG onset time of the two vasti muscles recorded with either method. In the case of studying the absolute EMG onset time, findings from the DP-P method cannot be directly compared with the mean + 3 s.d. method.

The difference between the two methods may be explained by the fact that the DP-P and mean + 3 s.d. methods have inherent difference in their constructs. The mean + 3 s.d. method relies on signal processing using a 50 Hz low-pass filter and full-wave rectifier to attenuate the raw EMG signals, so as to transform the AC signals to DC. These processing techniques would smoothen the signals substantially. Since the muscle onset time is the first instance when the signal exceeds the preset threshold, the relatively smoother waveform would have relatively lower threshold as it has proportionally smaller values of mean and standard deviation. In contrast, the DP-P method does not process the raw EMG signal, it only sets the threshold entirely based on the resting signal amplitude, thus the DP-P would take relatively longer time to reach the onset point than the mean + 3 s.d., which explains the 3 ms delay in the muscle onset detection with the DP-P method.

3.7. Conclusion

This study shows that the DP-P method is a reliable method to detect the EMG onset of the vasti muscles in isometric knee extension. The within- and between-day reliability is particularly high at 50% of MVC. There is a constant delay of about 3 ms with the DP-P detection method as compared to the mean + 3 s.d. method, thus direct comparison of the absolute muscle onset time of these two methods should not be done.

3.8. Relevance to the main study

Having determined the reliability and comparability of the DP-P method to an established mean + 3 s.d. method of EMG onset detection for VMO and VL, the simplicity of the DP-P is a bonus for handling large amount of EMG data. This method will be used in the main study of this thesis (Chapter 6) that examines the effects of weight training on the potential adaptation of the vasti muscles.

Chapter 4

(Study III)

Quantification of Lateral Patellar Mobility

4.1. Background and literature review

Increase in lateral patellar mobility has been identified as a predisposing factor for patellofemoral pain syndrome (PFPS) in both cross-sectional and prospective studies (Kujala et al 1986; Witvrouw et al 2000). Patellar hypermobility or instability could be a consequence of patellar subluxation or dislocation, which, if not managed properly, will predispose the subject to develop patellofemoral pain (Hughston et al 1984; LaBella 2004). Instability of the patellofemoral joint can be defined as the patella being inherently unstable or unable to maintain normal tracking or equilibrium in response to external perturbations (Padua & Blackburn 2003).

Clinically, physicians or physical therapists can perform a manual patellar gliding test to estimate the patellar mobility by positioning a patient in supine lying with knee extended, and moving the patella medially or laterally. If the extent of displacement is more than 75% of the width of patella, it is considered hyper-mobile (Fulkerson 2002). Since a more precise measurement of passive patellar mobility would be desirable for diagnostic and research purposes, a few clinical and laboratory testing methods have been developed (Minkoff & Fein 1989; Fithian et al 1995; Teitge et al 1996).

Radiographically, Minkoff and Fein (1989) and Teitge et al (1996) positioned the patients on inclined supine lying with knees flexed at 30°; the examiner used a handheld force gauge (Patellar Pusher, MedMetric, San Diego, CA, USA) to displace the patella medially or laterally so that the extent of patellar displacement can be viewed and calculated on x-ray films (fig 4-1). This stress radiograph of patellofemoral joint was not widely used in clinical settings, and was more of a research tool (Deie et al 2005). Also, this method has certain limitations, the x-ray image must be taken with the knees flexed at 30° or more, otherwise the image of patellofemoral joint would be blocked by the proximal tibia (Minkoff & Fein 1989). However, the patellofemoral joint is most mobile with a fully extended knee (Fox 1975), the patellofemoral articular contact area is relatively greater with knee flexion, thus the radiographic examination with a flexed knee is not ideal if the soft tissues of patellofemoral joint are evaluated for their elasticity or stiffness (Desio et al 1998; Nomura et al 2006).

Another limitation of the above method is that the lateral patellar displacing force cannot be directed purely horizontally due to the presence of the other knee that would block the horizontal pathway of the mediolateral gliding. Therefore, the patella will be shifted laterally and tilted transversely (fig 4-2).

With a more simple method than stress radiograph, Fithian et al (1995) reported using a handheld dynamometer to standardize the patellar translating force and a linear displacement sensor to measure the passive patellar mobility at 30° knee flexion (fig 4-3).

However, all the above methods use a handheld dynamometer which relies heavily on the examiner’s eye-hand coordination in controlling the patellar displacement by holding the dynamometer steadily and perpendicularly to the edge of patella. Thus, these methods are prone to errors which can significantly affect the measurement accuracy, because the passive range of side-way patellar mobility is usually less than 20mm in asymptomatic subjects (Fox 1975).

Figure 4-1: Stress radiograph of patellofemroal joint. Teitge et al (1996) with permission.

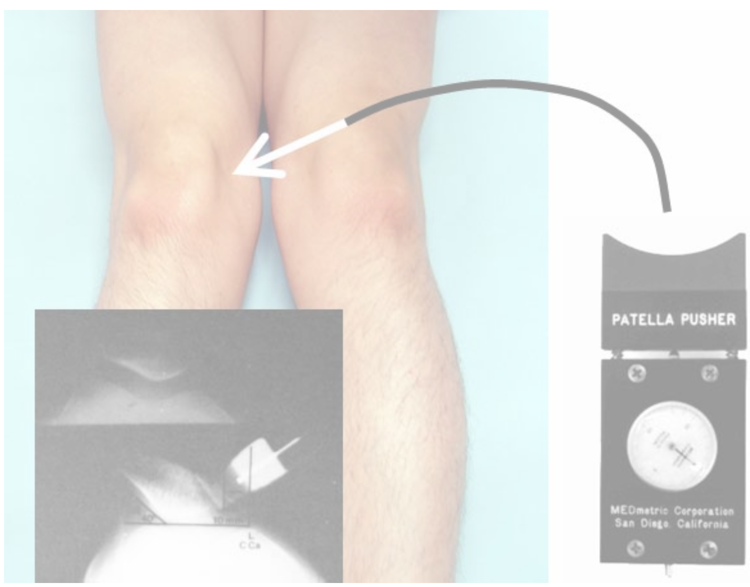


Figure 4-2: Axial stress x-ray of patellofemoral joint. The patella being displaced by a force dynamometer (patella pusher), the displacing force was in both horizontal and downward directions. X-ray image from Minkoff & Fein (1989) with permission.

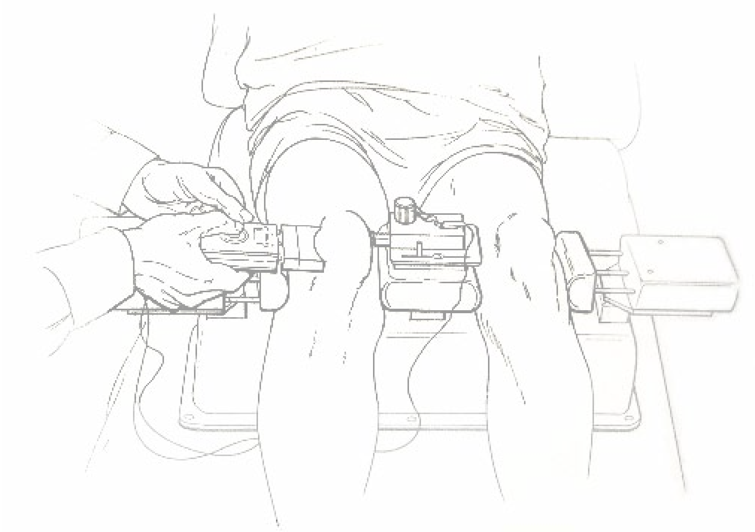


Figure 4-3: Passive patellar gliding measurement. Fithian et al (1995) with permission.

4.2. Objectives

In order to reliably quantify the lateral patellar mobility, and to examine the correlations between patellar mobility and geometry of the patellofemoral joint, this study aims to:

1) develop a mechanical method to quantify passive patellar gliding which can be used in the main study of this thesis (Chapter 6),

2) record the axial and sagittal views of the knee with magnetic resonance imaging (MRI),

3) correlate the results of MRI and mechanical patellar gliding, and

4) define the clinical significance of these measurements.

The work of the current study was presented at the 2nd Asia Pacific Conference on Biomechanics, 2005, Taipei, Taiwan and the Annual Meeting of the American Society of Biomechanics, 2006, Blacksburg, VA, USA, and will be published in American Journal of Physical Medicine and Rehabilitation (Appendices 9, 10, 11)

4.3. Hypothesis

1) Newly designed equipment can reliably measure the passive lateral patellar mobility.
2) The magnitude of patellar mobility can be correlated to the geometries of patellofemoral joint obtained from the MRI examination.

4.4. Methods

<u>Subjects</u>

1st group

Thirteen able-bodied volunteers (11 males, 2 females) aged between 21 and 40 years (mean ± SD = 26.6 ± 4.7) were recruited for investigating the reliability of instrumented patellar gliding test.

2nd group

Seventeen able-bodied volunteers (13 males, 4 females) aged between 18 and 35 years (mean ± SD = 26.7 ± 5.1) were recruited for investigating of the relationships between the geometrical features of patellofemoral joint and lateral patellar mobility.

The study was approved by the Human Ethics Sub-committee of the Hong Kong Polytechnic University before data collection (Appendix 5). All subjects were requested to give their written consent prior to the study (Appendix 6).

<u>List of apparatus</u>

1) MRI scanner (Magnetom Avanto, Siemens AG, Erlangen, Germany).
2) Laser-cross projector (SL01, Land, Hong Kong).
3) Adhesive dressing membrane (Tegaderm, 3M Health Care, St. Paul, MN, USA).

4) Custom made instrument of lateral patellar gliding test, as an assembly with:

- force gauge transducer (AEF-5, Aikon Engineering Co., Osaka, Japan),
- magnetic motion tracker (pciBird, Ascension Technology, Burlington, VT, USA)
- adjustable platform (Poly-Jaque plastic lab jack, Scienceware, Pequannock, NJ, USA) (fig 4-4, 4-5).

5) Mechanomyographic (MMG) biofeedback instrument, as an assembly with:

- air coupled transducer (Pulse pickup, 21051D, Hewlett Packard, Palo Alto, CA, USA)
- EMG biofeedback unit (Myotrac, Thought Technology, Montreal, Quebec, Canada) (fig 4-6).

<u>Instrumented lateral patellar gliding test</u>

The test used a DC magnetic motion tracker with 0.5 mm accuracy (pciBIRD, Ascension Technology., Burlington, VT, USA) that consisted of three devices, namely magnetic field transmitter, magnetic field sensor (fig 4-4), and computer digital signal processing card. The tracker was controlled by an IBM compatible personal computer with a Pentium III-500Hz microprocessor (Intel, Santa Clara, CA, USA) running Windows 2000 operating system (Microsoft, Redmond, WA, USA). Additional software used was the Shared Input Devices Controller (Swiss Federal Institute of Technology, Lausanne, Switzerland) for collecting 3-axis ordinate data while the sensor was fixed on a manual patellar gliding apparatus (AEF-5, Aikon Engineering Co., Osaka, Japan) that seated onto a custom-made linear slide and supported by a height-adjustable stand (Poly-Jaque plastic lab jack, Scienceware, Pequannock, NJ, USA) (fig 4-4).

The magnetic field transmitter and the sensor were located lateral and medial to the tested knee respectively, thus the passive patellar motion can be detected while the patella was being pushed (fig 4-5). Data of the transverse axis were collected as the medial to lateral patellar motion was the primary interest in the measurement, and was collected at a sampling rate of 100Hz.

During the test, the subject was positioned in supine lying with knees extended and feet together. The feet and lower legs were secured with no active movement allowed (fig 4-5). A laser-cross projector (SL01, Land, Hong Kong) was mounted on a jig above the knees which emitted a cross mark on the plinth for standardizing the position of the subject (fig 4-4). The patellar glide apparatus fitted with the magnetic motion tracker was used to instrumentally glide the subject's patella laterally in a constant speed of approximately 2 cm/second with 15N force. The data of the patellar displacement was analyzed with Excel software (Office 2000, Microsoft, Redmond, WA, USA).

Both knees were tested separately with three continuous passive gliding movements and the average of the three recordings was calculated to minimize random error. In order to help the subjects to relax their quadriceps muscles during the test, real-time biofeedback signal was provided with mechanomyography (MMG) of the medial quadriceps muscles. The MMG signal was detected via an air coupled transducer (Pulse pickup, 21051D, Hewlett Packard, Palo Alto, CA, USA) connected to a biofeedback unit (Myotrac, Thought Technology, Montreal, Quebec, Canada) which would provide an audio alarm when the quadriceps' MMG signal (root mean square, 100ms time constant) was over its resting threshold (Bolton et al 1989; Evetovich et al 2007). The MMG transducer was fixed on the medial quadriceps with an adhesive dressing membrane (Tegaderm, 3M Health Care, St. Paul, MN, USA) (fig 4-6).

The 1st group of subjects (n = 13) were tested twice with one-week apart for examining the test-retest reliability of the instrumented patellar gliding test.

The 2nd group of subjects (n = 17) were tested once with the patellar gliding test and then scanned with MRI for their knees and thighs.

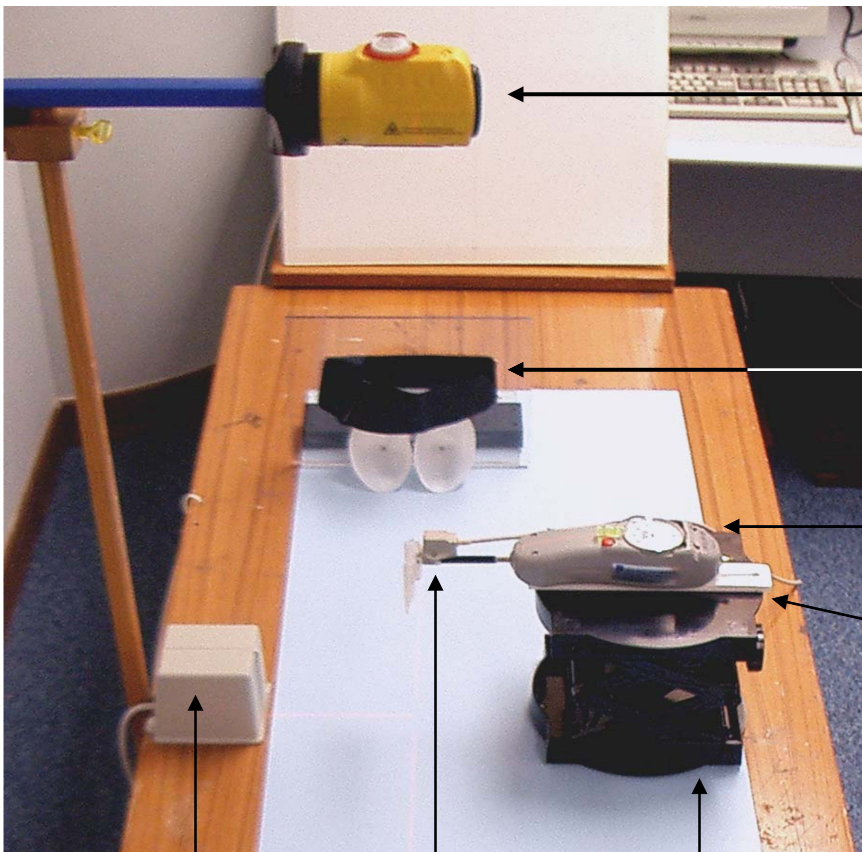


Fig 4-4: Instruments for lateral patellar gliding test.

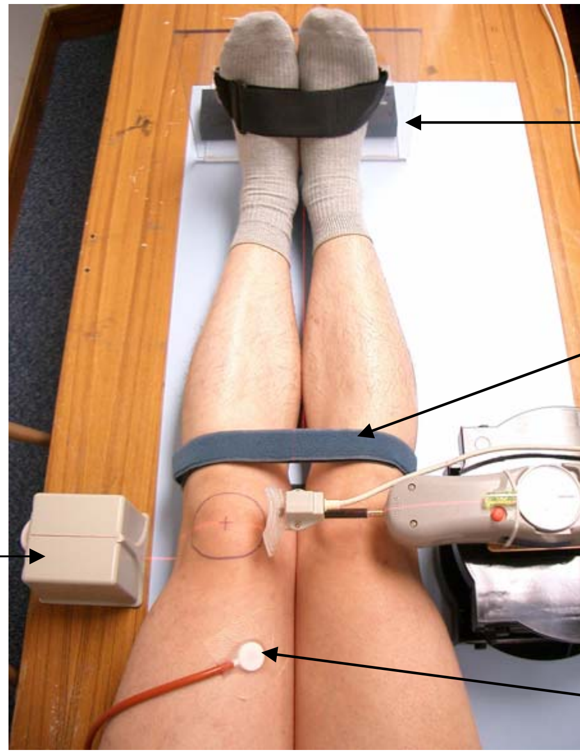


Figure 4-5: Instrumented lateral patellar gliding test in use.

Figure 4-6: MMG biofeedback instrument.

MRI

After the patellar gliding test, the 2nd group of subjects received MR imaging of their knees and lower legs. A 1.5 Tesla MRI scanner with surface coils (Magnetom Avanto, Siemens AG, Erlangen, Germany) was used to record the sagittal (T1, TI: 600ms, TR: 1180ms, TE: 4.32ms, slice thickness: 2mm) and axial views (T1, TR: 420ms, TE: 50ms, slice thickness: 3mm) of the subjects' knees and lower thighs with the subjects in the same positioning as the patellar gliding test. Foam inserts were used to support the surface coils in order to keep the anterior knees from being compressed by the coils (fig 4-7). The 2-mm and 3-mm MRI slices were chosen because it can provide clinically acceptable spatial resolution and this thickness has been used conventionally for the examination of knee ligaments and tibiofemoral cartilage injuries (Cicuttini et al 2004). Image datasets were saved in DICOM format for off-line analyses.

Off-line analyses were done with medical image analysis software (MIPAV, version 2.0, National Institutes of Health, Bethesda, MA, USA) and a graphic digitizer (CTE-430, Wacom, Vancouver, WA, USA). The MR images were analyzed for the patellar tendon length/patellar length (TL/PL) ratio (Miller et al 1996; Shabshin et al 2004), the lateral trochlear inclination (Carrillon et al 2000) and the biomechanical inclination angle (fig 4-8, 4-9, 4-10).

Figure 4-7: Positioning of subjects for the MRI.

Figure 4-8: Patellar tendon length/patellar length (TL/PL) ratio. The ratio was formed by the patellar tendon length divided by the diagonal patellar length at the sagittal image that demonstrated maximal patellar length.

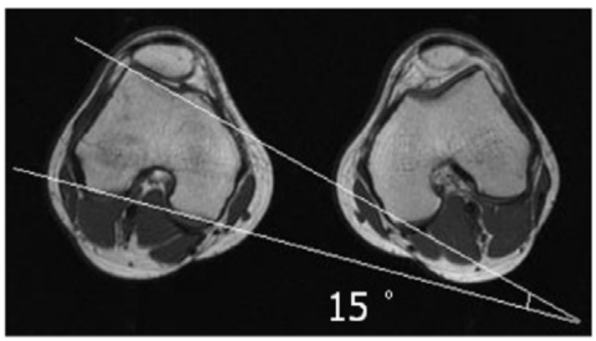


Figure 4-9: Lateral trochlear inclination. The angle was subtended by a line tangential to the lateral trochlear facet and the posterior condylar reference line on the first caniocaudal image that demonstrated cartlilaginous trochlea.

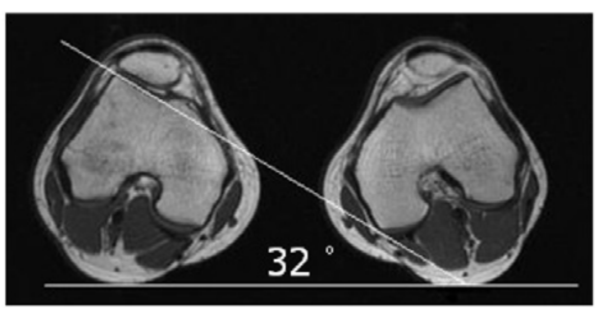


Figure 4-10: Biomechanical inclination angle. The angle was subtended by a line tangential to the lateral trochlear facet and the supporting surface.

Statistical analysis

Test-retest reliability of the instrumented lateral patellar gliding test: Paired data of day 1 and day 2 regarding left leg, right leg and pooled left-right legs were respectively analyzed using intraclass correlation coefficient (ICC (3,1), p = 0.05, consistency).

Correlation between MRI based measurements and passive lateral patellar displacement: Since both legs of each subject (n = 17) were measured, this resulted in 34 data. Pearson's coefficient (p = 0.05, two-tailed) was calculated to examine the correlations based on data of left leg, right leg, and pooled left-right legs respectively.

4.5. Results

For intra-rater reliability of the patellar gliding test with a 7-day interval, the ICC (3,1) values were 0.97 (95% CI: 0.90 - 0.99) for left leg, 0.94 (95% CI: 0.82 - 0.98) for right leg, 0.96 (95% CI: 0.94 - 0.98) for pooled left-right legs.

Table 4-1 presents the results of the descriptive data of the patellar gliding test and MRI measurements.

Table 4-1: Descriptive data of patellar gliding test and MRI based measurements.

Parameter	n	Mean	SD	Range
Patellar gliding test (mm)	34	14.2	4	10-25
TL/PL ratio	34	0.95	0.14	0.74-1.33
Lateral trochlear inclination (deg)	34	19.4	3.9	14-27
Biomechanical inclination angle (deg)	34	23	3.5	14-28

The correlations between the patellar gliding test and the MRI based measurements were listed in the table 4-2.

Table 4-2: Correlation between lateral patellar gliding and MRI based measurements.

Person's correlation	Left leg		Right leg		Pooled left-right	
	r	p	r	p	r	p
TL/PL ratio – Patellar gliding test	0.29	0.246	0.33	0.192	0.31	0.073
Lat. trochlear inclination – Patellar gliding test	-0.49	0.042 *	-0.57	0.016 *	-0.53	0.001*
Biomechanical inclination – Patellar gliding test	-0.68	0.002 *	-0.65	0.004 *	-0.67	0.001*

* = statistical significance

The graphical presentation of the correlations between passive lateral patellar mobility and lateral trochlear inclination, and biomechanical inclination angle for the pooled left-right legs data was shown in figure 4-11.

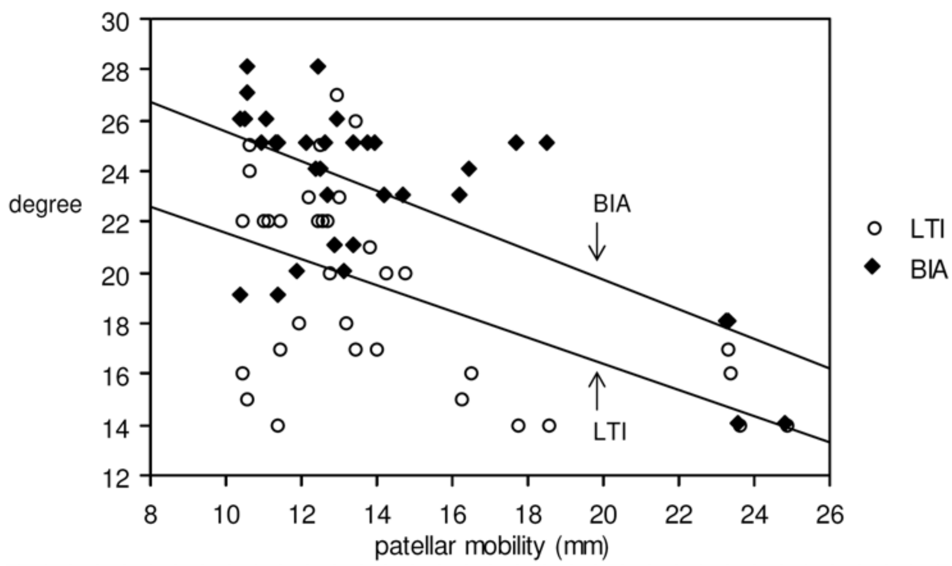


Figure 4-11: Bivariate plot with linear regression lines. The plot showed relationships between patellar mobility and MRI based measurements.

LTI = lateral trochlear inclination, BIA = biomechanical inclination angle

4.6. Discussion

Carrillon et al (2000) classified that a lateral trochlear inclination of ≤ 11° as femoral groove dysplasia. This small lateral trochlear inclination would have significant association with patellar instability. Shabshin et al (2004) studied 245 knees with MR imaging and reported that a TL/PL ratio of between 0.74 and 1.5 to be normal. If the ratio is over 1.5, it is classified as patellar alta and this implies a potentially unstable patella due to the excessively long patellar tendon. Based on these criteria, all the subjects in this study would be regarded as within the norm.

The current findings suggests that the lateral passive mobility of patella was weakly correlated to the length of patellar tendon (TL/PL ratio), but it was significantly correlated with the femoral trochlea in terms of the lateral trochlear inclination and the biomechanical inclination angle (table 4.2, fig 4-11). As these angles became smaller, the passive patellar mobility tended to increase. The values of correlation suggest that the biomechanical inclination angle to be more predictive of the passive patellar mobility than the lateral trochlear inclination. The findings also suggest that the patellar gliding is a simple and valid clinical assessment for estimating the femoral trochlear geometry.

The lateral trochlear inclination would not be affected by the hip rotation angle or the angle of femoral torsion (Hoiseth et al 1989), but the biomechanical inclination angle would be affected by these angles. This point is clinically important because when a clinician performs the patellar gliding test (manual or instrumented) on a patient, the hip rotation angle (limb position) must be controlled as it can affect the biomechanical inclination angle that may alter the magnitude of the lateral patellar displacement.

It is believed that when the patella was glided laterally, the gliding force was directed horizontally and diagonally downwards due to the irregular shape of patella. The force had induced a reaction force on the lateral facet of the trochlear groove, and this reaction generated friction in the lateral patellofemoral joint that decreased the patellar translation (fig 4-12). In other words, an increase in the biomechanical inclination angle would hinder the patellar lateral gliding movement; and a decrease of it would facilitate the movement.

The patellar gliding test was only performed for the lateral side since the majority of the patellar disorders happen on the lateral side (Hughston et al 1984). Nonweiler & DeLee (1994) and Fulkerson (2004) reported that medial patellar subluxation is clinically uncommon. If the subluxation presents, it is always a complication of lateral retinacular release. Also, Cutbill et al (1997), Thomee et al (1999) and LaBotz (2004) stated that excessive lateral patellar tracking plays a role in the development of PFPS. Therefore, establishing patellar stability on the lateral side is clinically important.

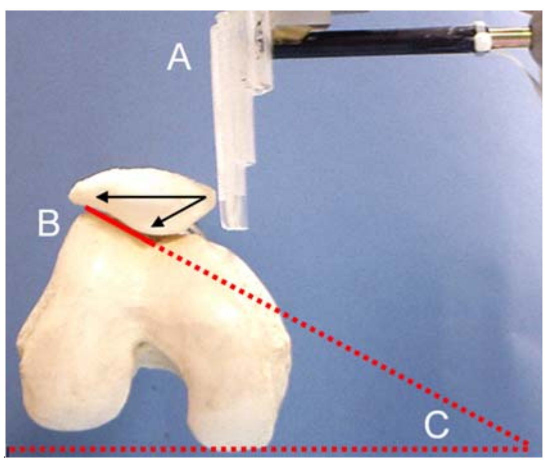


Figure 4-12: Horizontal gliding force created a reaction force in the patellofemoral joint. (A) = gliding force; (B) = reaction force in the lateral patellofemoral joint. The reaction force may increase or decrease with the biomechanical inclination angle (C) as well as hip rotation angle.

The design of the current instrumented patellar gliding test has been proved to be repeatable, and it has also solved the problems of the previous versions of passive lateral patellar mobility measurement that the lateral patellar translation was not purely horizontal (Minkoff & Fein 1989; Teitge et al 1996), and the translation direction heavily relied on the examiner's eye-hand co-ordination (Minkoff & Fein 1989; Fithian et al 1995; Teitge et al 1996). The L-shape patellar pusher in this study could cross over the contralateral knee and push the tested patella in a horizontal direction (fig 4-13), and the force gauge was guided by a linear bearing with the laser-cross marking which enabled the standardization of the subjects' position (fig 4-4, 4-5). These design features have improved the accuracy of the patellar gliding test and the logistics of which does not require highly sophisticated set-up.

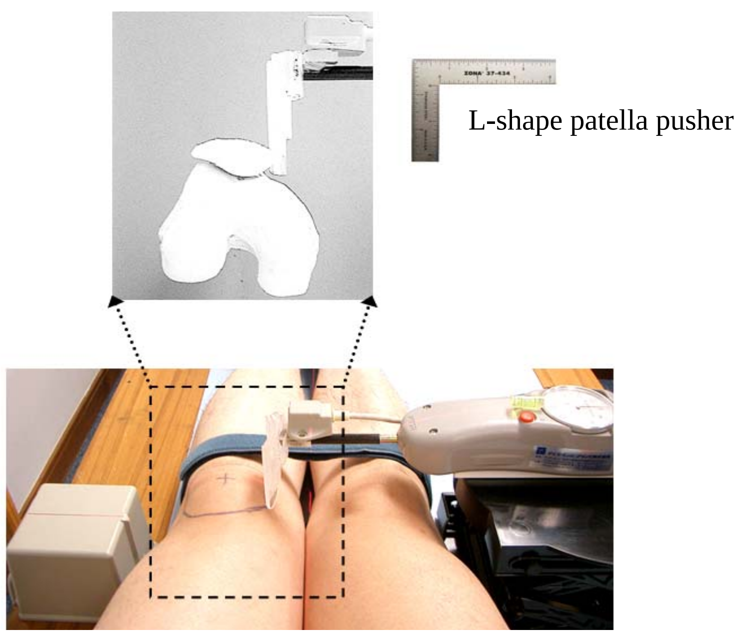


Figure 4-13: Patellar gliding was operated in medio-lateral and horizontal direction.

However, there are some limitations in this non-blinded study that need to be addressed. Only 17 subjects (34 knees) were tested with most of them being males (n = 13); the potential gender difference in lower limb alignments or soft tissue compliance may affect the femoral geometry-patellar mobility correlations because it has been reported that female subjects have a relatively higher TL/PL ratio (Shabshin et al 2004). Since the subjects being tested were all young and asymptomatic, the findings may not be directly generalized to older subjects or those with knee pathologies. In the patellar gliding test, a standard gliding force of 15 N was used rather than a percentage of the subjects' body weight, the potential effect of the non-individualized gliding force on the correlations is therefore unknown.

4.7. Conclusion

When clinicians treat people with excessive patellar mobility, knowing the magnitude and cause of the problem is essential for planning the treatment programs and evaluation of the treatment progress. While the magnitude and the cause of patellar mobility can be examined via mechanical and radiographic means, there was little information on the correlation between the mechanical and radiographic findings. It is concluded from the present study that the position of lateral trochlear groove is a determining factor that reflects the lateral patellar mobility rather than the patellar tendon length among asymptomatic subjects. Since the hip rotation angle can affect the position of lateral trochlear groove, clinicians who perform lateral patellar gliding test should recognize the hip angle as being critical and needs to be controlled.

4.8. Relevance to the main study

The patellar gliding instrument developed in this study will be used in Chapter 6 in which the potential changes in passive lateral patellar mobility of the subjects with and without weight training are evaluated.

Chapter 5

(Study IV, V)

Imaging for Quantifying Patellar Position and Vastus Medialis Obliquus Size

Study IV

Validity of magnetic resonance imaging for patellar tilt angle

5.1.1. Background and literature review

The etiology of patellofemoral pain syndrome (PFPS) is multi-factorial, and certain contributing factors such as excessive patellar tilt can present with subtle physical signs that are difficult to be quantified or even detected by physical examinations in the clinical settings. Although a few clinical methods using manual palpation and tape measurement for estimating the extent of patellar tilt were developed (McConnell 1996; Ingersoll 2000), validity and reliability of these methods are yet to be proved (Watson et al 2001; McEwan et al 2007). Conversely, medical imaging including x-ray, computed tomography (CT) and magnetic resonance image (MRI) have been regarded as objective and reliable methods for quantifying the angle of patellar tilting (Beaconsfield et al 1994; Murray et al 1999; Muhle et al 1999; Elias & White 2004).

The excessive lateral patellar tilt has been associated with increased lateral patellofemoral joint pressure that stresses the retro-patellar subchondral bone (Hungerford & Maureen 1979; Draper et al 2007) and causes a localized anterior knee pain (Cutbill et al 1997; Thomee et al 1999; LaBotz 2004), therefore this condition is also known as lateral patellar compression syndrome (Larson et al 1979; Doucette & Goble 1992) (fig 5-1).

Kujala et al (1993) reported a positive correlation between the lateral patellar tilt by MRI measurement in patients with anterior knee pain and their subjective symptoms. Pookarnjanamorakot et al (1998) analyzed the degree of lateral patella tilt with MRI and its correlation with the symptoms of anterior knee pain, and found that the average patella tilt angle (fig 5-2) for those without anterior knee pain was 12.8° (SD = 8.4); whereas it was 6.3° (SD = 3.9) for persons with anterior knee pain. However, both studies were retrospective in nature and therefore it could not determine whether the increased patellar tilt was a predisposing factor or a result of anterior knee pain.

Since the patella is minimally constrained and its position is largely determined by the surrounding soft tissues at full knee extension (Fulkerson & Gossling 1980), the optimal position for imaging the patellar tilt is with an axial view at full knee extension (Minkoff & Fein 1989; Vahasarja et al 1996). When the knee is flexed by more than 20°, the patella is drawn into the trochlear sulcus due to the closing angle between the quadriceps and patellar tendon in the sagittal plane, hence patellar stability and patellofemoral joint contact area increase with knee flexion (Hungerford & Maureen 1979; Hinterwimmer et al 2004). However, axial x-ray of patellofemoral joint, also known as Merchant or sunrise view, cannot produce the most ideal image of static patellar orientation since the positioning requires the knee to flex by at least 30° (fig 5-3), otherwise the patellofemoral image would be blocked by the superimposing proximal tibia (Minkoff & Fein 1989; Vahasarja et al 1996).

Technically, only MRI and CT are able to capture the axial images of the cross-section of a patellofemoral joint with knee extended. Thus, both modalities have been extensively used by physicians and researchers in investigating the angle of patellar tilt in subjects with or without patellofemoral pain (Inoue et al 1988; Conway et al 1991; Beaconsfield et al 1994; Guzzanti et al 1994; Delgado-Martinez et al 1996; Powers et al 1999, 2004; Witonski & Goraj 1999; Sathe et al 2002; Ward et al 2002).

There were two image-based measurements using lateral patellar facet as a reference for calculating the degree of patellar tilt at mid-transverse patellar level, known as patellar tilt angle (PTA) (fig 5-2) and lateral patellofemoral angle (LPFA) (fig 5-4) (Delgado-Martinez et al 2000; Shibanuma et al 2005). Wu and Shih (2004) reported both PTA and LPFA were highly correlated with each other (r = 0.99) and suitable for qualification of the lateral patellar tilt.

Furthermore, both measurements were reported to have high between-day reliability. Delgado-Martinez et al (2000) reported that the intra-rater reliability of both CT based LPFA and PTA measurement to be 0.95 of the Pearson's coefficient, whereas the intraclass correlation coefficient (ICC) for MRI based PTA measurement was reported to be 0.95 (Sathe et al 2002).

For the MRI based measurement of patellar tilt, the PTA was reported to be a more preferable method than the LPFA, since PTA was less sensitive to the location of the imaging plane as compared with LPFA. In the PTA measurement, the posterior condyles provided a stable reference line throughout a serial cranio-caudal imaging, while the LPFA uses the femoral trochlea as a reference, which could change craniocaudally (Fulkerson et al 1987; Shibanuma et al 2005).

Although MRI based PTA measurement has been regarded a repeatable indicator for the extent of lateral patellar tilt, little information is available on the validity of PTA measurement.

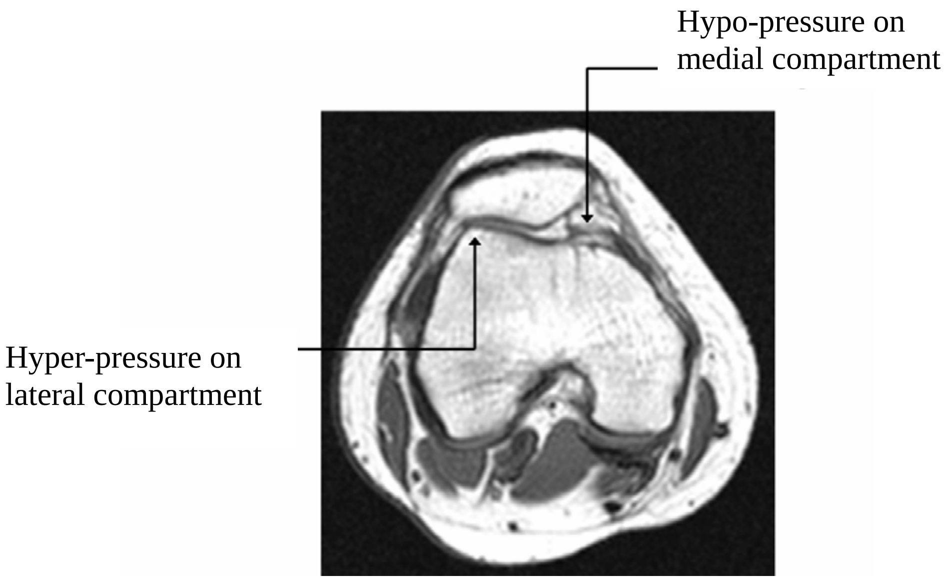


Figure 5-1: Right knee at inferior view illustrates the observable lateral patellar tilting. The excessive tit could cause extra-pressure on the lateral patellofemoral joint that stresses the subcondral bone, and reduced pressure on the medial side may lead to local malnutrition of the cartilage (Hungerford & Maureen 1979).

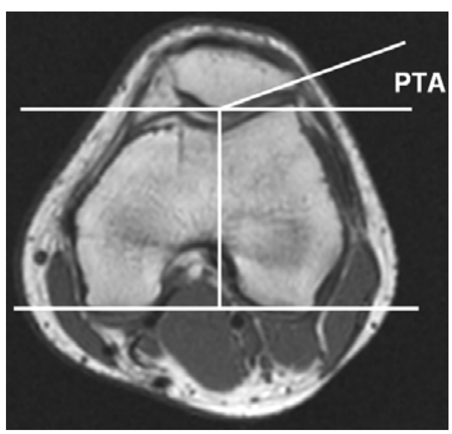


Figure 5-2: Patellar tilt angle (PTA). The angle is defined as an acute angle subtended by a line tangential to the lateral patellar facet, with a line parallel to the posterior condyles. The above is a left knee at inferior view.

Figure 5-3: Positioning for axial x-ray of patellofemoral joint. This view requires at least 30° of knee flexion (with permission from AliMed, Inc., Dedham, MA, USA).

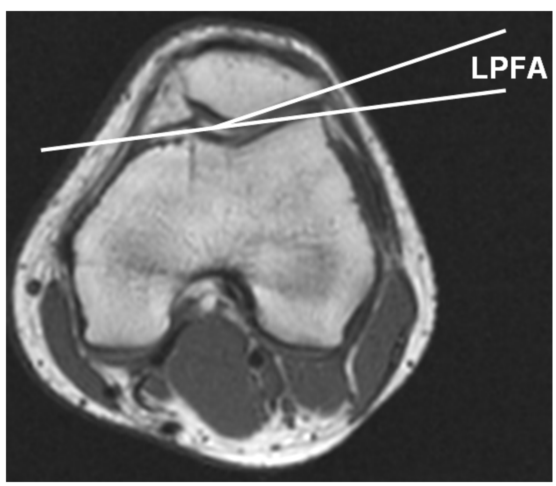


Figure 5-4: Lateral patellofemoral angle (LPFA). The angle is defined as an acute angle between a line provided by the femoral trochlea and the lateral patellar facet. The above is a knee at inferior view.

5.1.2. Objectives of study IV

This study aimed to compare the PTA measured directly on a frozen patellofemoral joint section of a pig's knee with the value obtained from MR imaging of the same specimen.

5.1.3. Methods

<u>List of apparatus</u>

1) 1.5 Tesla MRI scanner (Magnetom Avanto, Siemens AG, Erlangen, Germany).
2) Goniometer (Lafayette Instrument, Lafayette, IN, USA).
3) Form, plastic bags and plastic box (fig 5-5).

<u>Specimen preparation</u>

A fresh leg (right side, human food-graded) of the domestic pig (Sus Scrofa Domestica) had been stocked in a refrigerator at -40°C for 12 hours before being mechanically cut to the slices of 1.5cm thickness. Only two slices that showed the patellofemoral joint were selected for the MRI scanning (fig 5-5). The two slices provided four surfaces of the patellofemoral joint at a transverse view and allowed a direct manual goniometric measurement on the joint using a standard plastic goniometer (Lafayette Instrument, Lafayette, IN, USA).

<u>MR scanning</u>

The selected specimens were enclosed in two air-sealed plastic bags and were placed inside three supporting foam blocks in a plastic box to mimic a human subject lying supine for MR scanning (fig 5-5). A 1.5 Tesla MRI scanner with surface coils (Magnetom Avanto, Siemens AG, Erlangen, Germany) was used to image the specimens. The MRI protocol was T1 weighted, TR/TE: 420ms/50ms, slice thickness of 3mm and inter-slice gap of 0.3mm. The MR images were saved in the DICOM format and PTA was measured using image analysis software (MIPAV, version 2.0, National Institutes of Health, Bethesda, MA, USA) (fig 5-6).

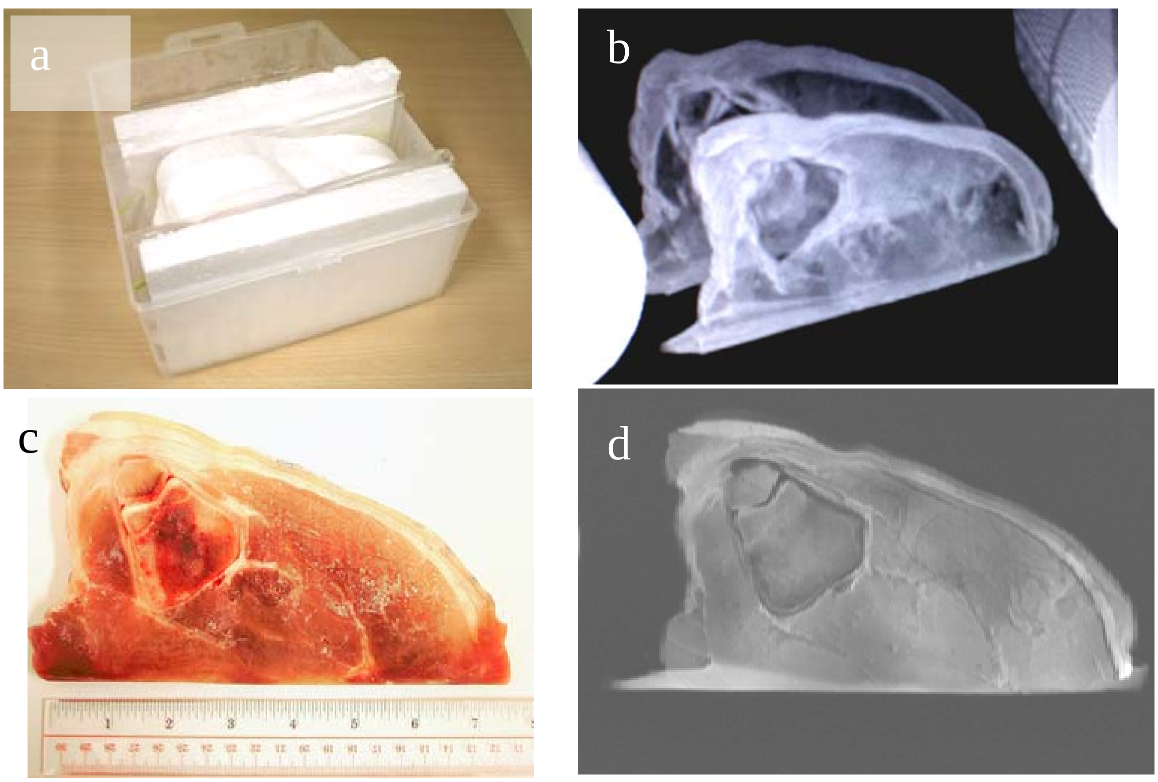


Figure 5-5: Setting of the MRI-validity study using a pig knee.

a = Specimens were placed vertical in a plastic box

b = 3D MR image of specimens corresponding to the picture a

c = Specimen that showed the cross-sectional area of patellofemoral joint

d = MR image of specimen corresponding to the picture c

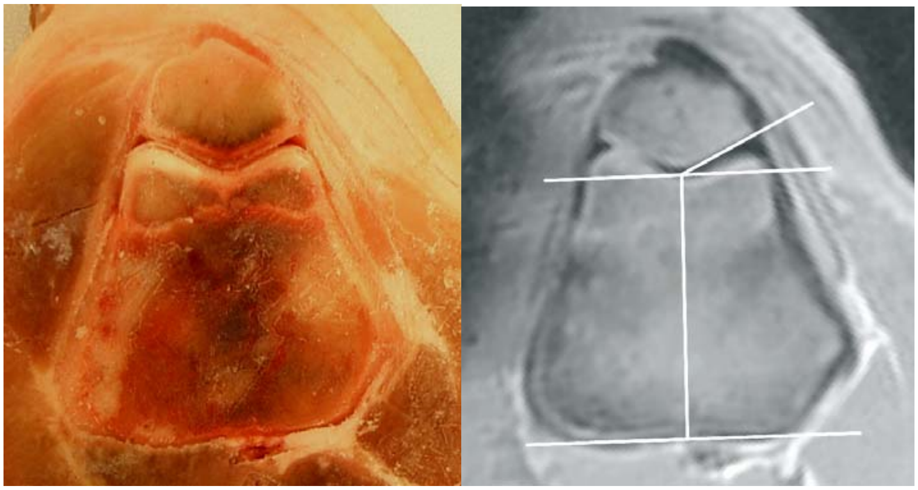


Figure 5-6: Actual picture and MR image of a pig patellofemoral joint.

<u>Data Analysis</u>

In light of this being a single sample study, absolute differences between the goniometric measured PTA with those obtained from MRI were calculated.

5.1.4. Results

The PTA measured in the specimens were 30°, 30°, 25° and 25° respectively, while the corresponding values obtained from the MR images were 31°, 30°, 25° and 26° respectively. The difference between the PTA measured on the specimens and the MR images was between 0 - 1°.

5.1.5. Discussion

Although MRI has been long regarded as a golden standard for visualizing mammal anatomical structures, MR images could give inaccurate shape and size for a given structure if the image with slice thickness greater than 1 mm (Port & Pomper 2000). However, only the latest and high-end MRI scanners can capture the 1-mm thick slices and usually the imaging is impractically time-consuming. The present study compared the PTA directly measured at a physical model and the PTA measured at the corresponding MR image with 3-mm thickness slice, and did not found the PTAs were significantly different in the images and the real model.

The model used in this study is a pig knee, the result should be transferable to human subjects because all components of the patellofemoral joint appear to be similar in both pig and human (Erne et al 2005). In view of high comparability between the goniometric measurements with the MRI measurements, current protocol (T1 weighted, TR/TE: 420ms/50ms, 3 mm slice thickness, inter-slice gap 0.3mm, field of view 35×20cm, matrix 448×256) of MR imaging is considered to be a valid tool to examine the PTA.

A 3-mm MRI slice was chosen for the PTA because it can provide clinically acceptable spatial resolution and this thickness has been conventionally used for the examination of knee ligaments and tibiofemoral cartilage injuries (Cicuttini et al 2004). If a thicker MRI slice thickness was chosen, the scanning time can be reduced, but the spatial resolution will decrease which would make the goniometric measurement on the MR images more susceptible to error (Shih et al 1993).

Because the MR scanner used in this study was located in a public hospital in which the regulation was set to limit the non-human scanning, and also the hygiene was a concern even the specimen was sealed, the number of specimen tested could not be increased.

Study V

Comparison of ultrasound and MRI for VMO measurement

5.2.1. Background and literature review

The measurement of human skeletal muscle cross-sectional area is important for assessing the effects of physical training, disuse, aging, diet or medication (Hubal et al 2005; Kanehisa et al 2005). Atrophy of vastus medialis obliquus (VMO) muscle is relatively common among patients with patellofemoral pain syndrome (Fox 1975; Doxey 1987; McConnell 1996; Dixit et al 2007). In order to monitor the effectiveness of the rehabilitation program for patellofemoral pain, clinicians may measure the distal circumference of the thigh with a measuring tape. However, the distal thigh contains VMO as well as other muscles and subcutaneous tissues, the circumference measurement may not precisely address the change in the VMO muscle.

MRI has been regarded as the gold standard for measuring muscle size because skeletal muscles cross-section area (CSA) measured from magnetic resonance image (MRI) is in good agreement with those obtained from cadaver sections, and the test-retest reliability with MRI is high (Fleckenstein et al 1991; Clague et al 1995). For vasti muscles, Beneke et al (1991) reported an average discrepancy of 1.2% between direct measurements and MRI based calculation on the cadavers. They also reported that MR imaging of thirty-eight subjects with mixed genders, the test-retest reliability of the quadriceps CSA measurement was 0.981. Mitsiopoulos et al (1998) reported that MRI provided precise measurements (r = 0.99) of skeletal muscle CSA throughout a wide range of values from 10-100cm^2 as compared with the corresponding cadaveric quadriceps. The authors also showed that both the inter-rater and intra-rater reliability of MR images of thigh muscles acquired on two separate days were 0.99.

In addition to MRI, human skeletal muscle CSA can also be assessed by diagnostic B-mode ultrasound. Miyatani et al (2000) investigated the validity of ultrasound for measuring the CSA of biceps and triceps in 26 young adults and compared with the corresponding MRI measurements. They reported that a correlation coefficient of 0.96 between the ultrasound measurements and that of the MRI. Esformes et al (2002) studied the muscle CSA of the tibialis anterior of six healthy subjects by comparing the ultrasound images with the corresponding MRI. The ultrasonic and MRI methods generated comparable results and the difference was calculated to be between 0.15% and 5.17%. The between-day reliability determined by intraclass correlation coefficient (ICC) for the ultrasound measurement was 0.99.

For ultrasound measurement of the quadriceps muscles, Reeves et al (2004) reported that the inter-day ICC value of the B-mode ultrasound for the CSA of vastus lateralis of six healthy adults was 0.99. When comparing the reading of the ultrasound measurement with the corresponding MRI, a correlation coefficient of 0.99 was found indicating good comparability of ultrasound with MRI. However, only the vastus lateralis muscle was measured in their study and the findings may not be directly applied to other muscles such as the VMO.

5.2.2. Objectives of study V

In view of fact that the correlation between the ultrasonic and MRI measurements on VMO muscle size has not been investigated, the present study aimed to:

1) design a hand-free ultrasound scanning method, and examine the validity of the

method for quantifying the VMO cross-sectional area by comparing it with corresponding MRI;

2) investigate between-day reliability of the above ultrasonic scanning for the VMO size.

The present study was presented at the 2007 annual conference of American Society of Biomechanics held at the San Francisco, CA, USA (Appendix 12).

5.2.3. Methods

Subjects

Six able-bodied subjects (4 males & 2 females, 18-35 years old) participated in this study on a voluntary basis. The study was approved by the Human Ethics Sub-committee of the Hong Kong Polytechnic University prior to data collection (Appendix 5). All subjects were asked to give their written consent (Appendix 6.1, 6.2) and refrain from vigorous physical activities for 48 hours before the study.

List of apparatus

1) 1.5 Tesla MRI scanner (Magnetom Avanto, Siemens AG, Erlangen, Germany).
2) Custom made ultrasound probe holder, as an assembly with:
   - lab jack (Avogadro's Lab Supply, Miller Placem, NY, USA)
   - tilt stage (SMS Optical Co., Hauppauge, NY, USA)
   - instrument clip (Stagebeat Ltd., Faenham, Surrey, UK) (fig 5-7).
3) B-mode ultrasound scanner with a 4 MHz annular array probe (Voyager, Ardent Sound, Mesa, AZ, USA) that was connected to an IBM-type computer with 1.7 G microprocessor (Pentium 4, Intel, Santa Clara, CA, USA) and Window XP (Home edition, Microsoft, Redmond, WA, USA) installed.

4) Ultrasound gel pad (Aquaflex, Parker Lab, Orange, NJ, USA).
5) MRI marker (MR-Spot, Bristol, CT, USA) (fig 5-8).
6) Laser-cross projector (SL01, Land, Hong Kong).
7) Surgical skin marker (Sklar Instruments, West Chester, PA, USA).
8) Distill water and spiller.

Ultrasound scanning

Subject's distal thigh at the suprapatellar level was horizontally marked with a surgical skin marker (Sklar Instruments, West Chester, PA, USA) to indicate the location of VMO (Nicholas et al 1976; Doxey 1987) (fig 5-8). While the subject was in supine lying with knees extended and feet together, the feet and lower legs were secured with no active movement allowed. A laser-cross projector (SL01, Land, Hong Kong) was mounted on a jig above the knees which emitted a cross mark on the bed for standardizing the position of the subjects (fig 5-9).

The ultrasound probe (Voyager, Ardent Sound, Mesa, AZ, USA) was totally supported by the custom-made probe holder that was adjustable in height and inclination angle (fig 5-10). A pre-molded ultrasound gel pad (Aquaflex, Parker Lab, Fairfield, NJ, USA) was used as an acoustic medium between the ultrasound probe and the distal thigh (fig 5-11). Small amount of distilled water was applied on the distal thigh for enhancing the contact between the gel pad and the skin. The cross-sectional images captured for VMO were stored in TIFF format for later analysis with the Kodak 1D 3.5 software (Kodak, Rochester, NY, USA) and manual tracing with a graphic digitizer (CTE-430, Wacom, Vancouver, WA, USA).

The twelve VMO muscles (6 subjects) were scanned again one week later. Thus, the test-retest reliability of the current ultrasound measurement setting could be examined.

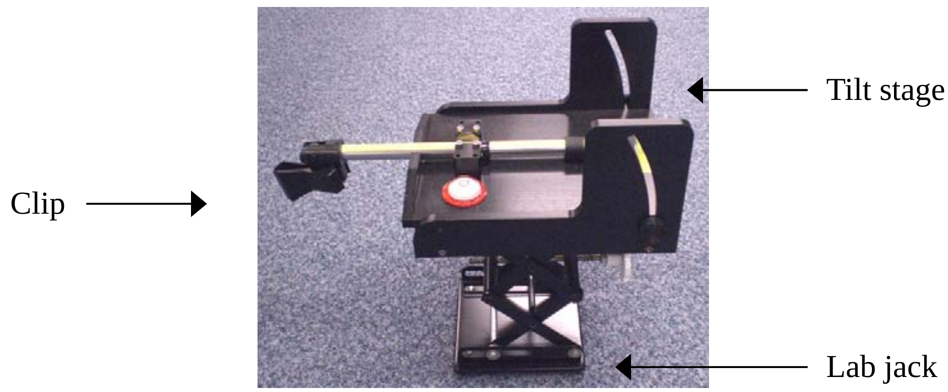


Figure 5-7: Hand-free ultrasound probe holder. It was designed to stabilize an ultrasound probe with minimal compression force on the scanned muscle.

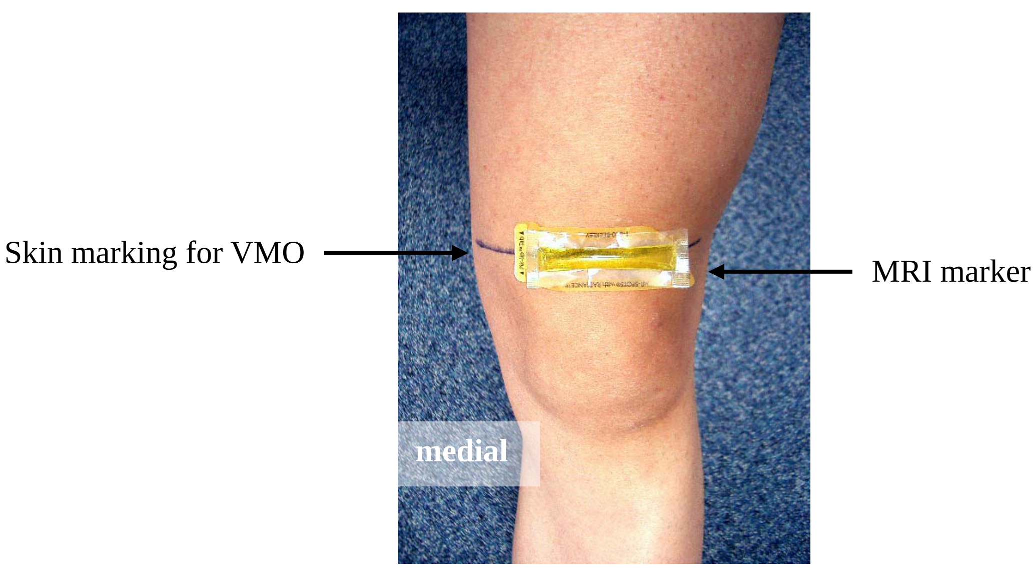


Figure 5.8: VMO muscle was marked for ultrasonic and MR imaging. The left thigh marked with a horizontal black line at the suprapatellar level for ultrasound scanning, while the MRI marker was placed at the same spot.

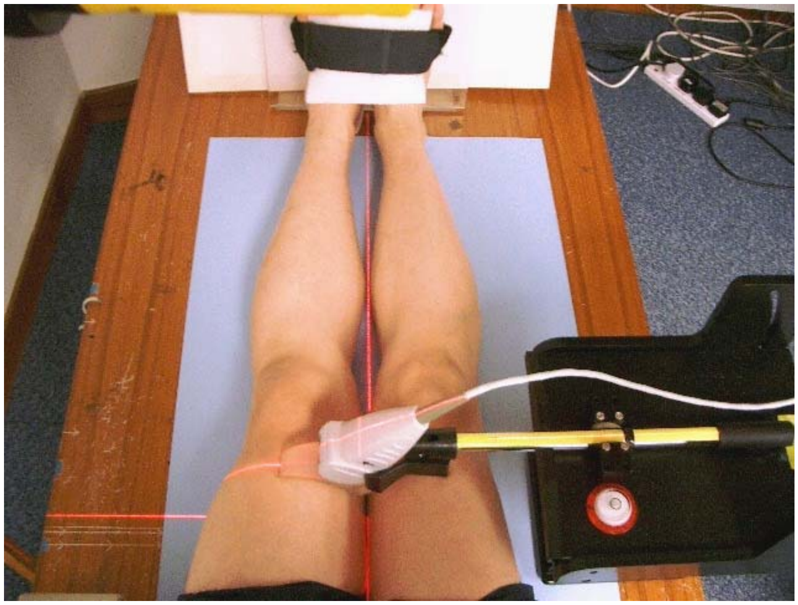


Figure 5-9: Setting of ultrasound scanning for VMO. A laser pointer projected a cross mark at the suprapatellar level and the distance between the sole of feet and the suprapatellar level was used to standardize repeatable positioning for the ultrasound scanning of the VMO.

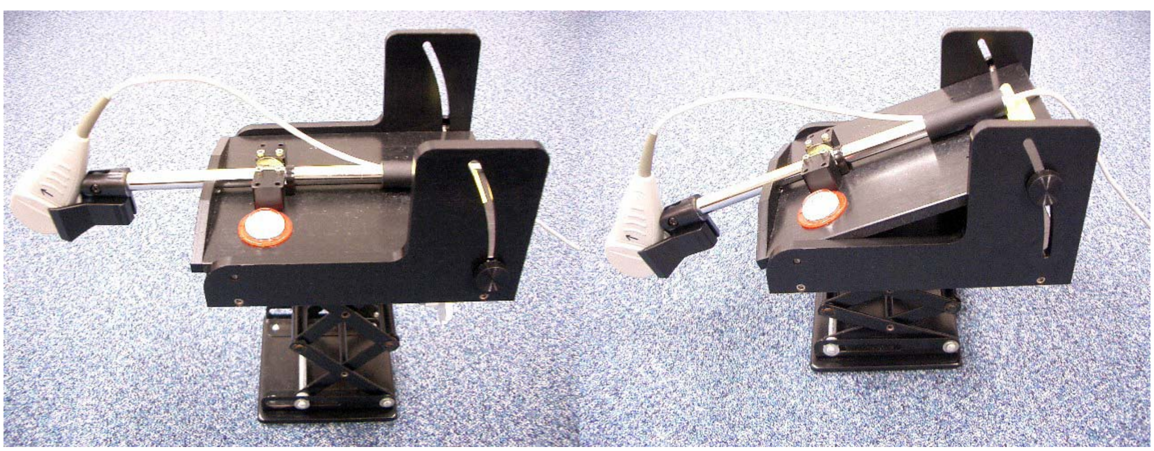


Figure 5-10: Static location and inclination angle of the ultrasound probe were adjustable. The axial VMO image can be captured with minimal compression on the top of the VMO.

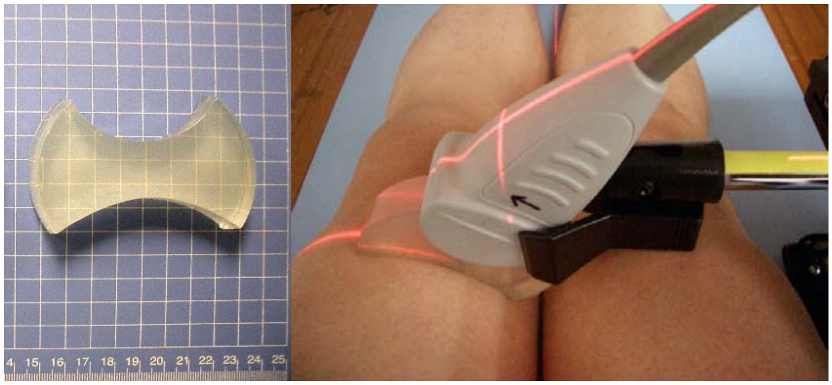


Figure 5-11: Ultrasonic scanning for VMO. The pre-molded ultrasound gel pad (left) acts as a medum between the ultrasound probe and the skin underneath (right).

<u>MRI scanning</u>

After the first ultrasound scanning, the subject received MR imaging of their thighs within 3 hours. A MRI marker (MR-Spot, Bristol, CT, USA) was placed on the suprapatellar level of thigh with the skin marking for the ultrasonic scan (fig 5-8). A 1.5 Tesla MRI scanner with surface coils (Magnetom Avanto, Siemens AG, Erlangen, Germany) was used to record the axial views (T1, TR: 420ms, TE: 50ms, slice thickness: 3mm) of the subjects' distal thighs with the subjects in the same positioning as in the ultrasound scanning. Foams were used to support the surface coils in order to keep the anterior thigh from being compressed by the coils. Image datasets were saved in DICOM format for an off-line analysis with image analysis software (MIPAV, version 2.0, National Institutes of Health, Bethesda, MA, USA) and a graphic digitizer (CTE-430, Wacom, Vancouver, WA, USA).

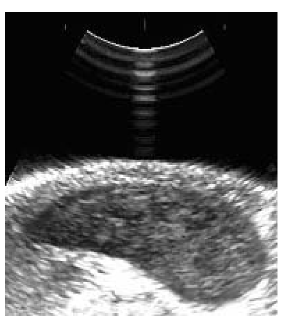
<u>Data analysis</u>

To find out the correlation between MRI and ultrasound based measurement on VMO cross-sectional area (CSA), Pearson's coefficient ($p = 0.05$) was calculated. Additionally, the means and standard deviations (SD) of differences between the values of two imaging methods were calculated based on the MRI data minus the corresponding ultrasound data and they were expressed as a % of the MRI data.

For test-retest reliability of ultrasound measurement of VMO size, ICC (3,1) (consistency, CI 95%, p = 0.05) was calculated for the consistency of the two sets of data on VMO CSA recorded with 7 days apart.

5.2.4. Results

The current study with twelve VMO muscles (six volunteers) revealed that the correlation for the VMO muscle measurement between MRI and ultrasound was significantly high (Pearson’s coefficient = 0.97), and the mean difference between these two methods the MRI was 5.7% (SD = 3%) with the ultrasound measurement slightly under estimating the VMO muscle as compared to the MRI (fig 5.12). Furthermore, the between-day ultrasound measurement was highly repeatable (ICC (3,1) = 0.96, 95% CI = 0.87 - 0.98).



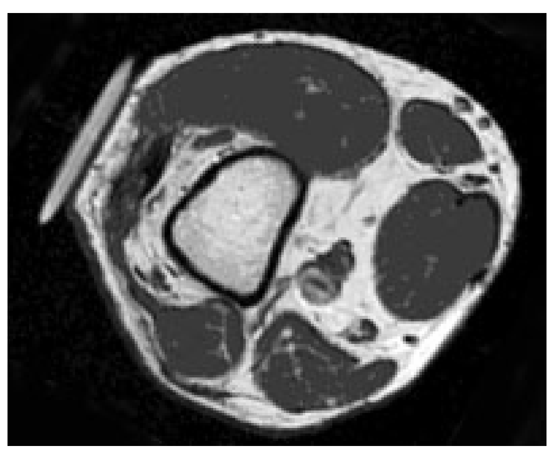


Figure 5-12: VMO in ultrasound and corresponding MRI image.

5.2.5. Discussion

The twelve VMO muscles from six subjects each provided two related samples for the MRI and repeated ultrasonic imaging respectively; the related data were used in data analyses since the size of left and right legs are always not the same. Also, there was room for error in measuring VMO size by ultrasound, such as imperfect positioning of the gel pad that deforms the VMO underneath or the manual tracking over/under outlining the muscles in calculating the cross sections. This means that an accurate measurement completed on one's left leg does not guarantee the same level of accuracy being made on the right side or vise versus.

Results of this study reveal that the current hand-free ultrasound measurement is comparable with the MRI based measurement for VMO size, and that the between-day ultrasound measurement is highly repeatable.

When compared to the MRI measurement, the ultrasound method under estimated the VMO muscle cross-sectional area by 5.7% (SD = 3%). This could be due to the ultrasound pad deforming the VMO underneath the skin. However, if the procedures are consistent for periodical measurements of the same muscle, the predictable under-estimation should not affect the meaningfulness and reliability of the ultrasound measurement.

The MRI is a costly diagnostic modality with certain contraindications. Subjects with claustrophobia, metal debris, cardiac or orthopedic implants inside their bodies are not suitable for MRI. In contrast, the B-mode ultrasound is without any known contraindications and side effect; it is widely available in hospitals and clinics worldwide. Thus, this modality should be regarded as a practical and an inexpensive method for quantification of the VMO size.

5.2.6. Relevance to the main study

The two studies in this chapter were pilot tests to confirm that both MR and ultrasound imaging methods are valid and reliable tools for quantification of the patellar tilt angle and the VMO size. These methods are integral parts of the key measurements for the main study (Chapter 6) that investigates the effect of weight training on skeletal muscles and soft tissues surrounding the patellofemoral joint.

Chapter 6

(Study VI)

Effect of Weight Training on the Active and Passive Patellar Stabilizers

6.1. Background and literature review

Non-traumatic anterior knee pain, or patellofemoral pain syndrome (PFPS), is one of the most common musculoskeletal conditions in both the general and athletic populations. The prevalence is between 10%-19%, with female athletes being more susceptible to this problem (Taunton et al 2002; Murray et al 2005). When exacerbated, PFPS could restrict the training of the athletes or even cease it altogether. Therefore, prevention of PFPS is vital in enhancing sports performance and prolonging athletes' careers.

Various theories have been proposed for the etiology of PFPS. One of the most prevailing views is that patellar maltracking or instability with or without overuse of the patellofemoral joint can stress the subchondral bone and patellar retinaculum (Fulkerson 2002; LaBella 2004; O'Donnell et al 2005). Instability of patella can be defined as the patellofemoral joint being inherently unstable or unable to maintain equilibrium in response to an external perturbation (Padua & Blackburn 2003).

Patellar stability is maintained by the interactions of active and passive patellar stabilizers (Arendt 2005). Vastus medialis obliquus (VMO) is regarded as the primary active stabilizer (Mariani & Caruso 1979; Souza & Gross 1991; Boucher et al 1992), while medial patellofemoral ligament (MPFL) is the primary passive stabilizer (Arendt et al 2002; Amis et al 2003; Hinton & Sharma 2003). These structures guard against excessive lateral patellar tracking, which is regarded as the underlying pathological mechanism of PFPS, patellar subluxation and dislocation (Hughston et al 1984; Fisher 1986; Wilk et al 1998; Grelsamer 2000). Theoretically, loading induced by regular resistance training can lead to morphological, biomechanical and neurological adaptations of the patellar stabilizers (Hayashi 1996; Folland & Williams 2007), which in turn, may enhance the patellar stability and minimize the risk of developing PFPS.

It has long been reported that physical exercise may well lead to hypertrophy of the skeletal muscles and strengthening of the ligaments which are important for sports injury prevention (Reid & Schiffbauer 1957; Adams 1966). A number of researchers have studied and reported the effects of physical exercise on injury prevention and rehabilitation. The following discussion presents an outline of the findings. Cahill and Griffith (1978) reported a decrease in the incidence and severity of knee injuries in high school male football players after a 6-week pre-season conditioning that included weight training, agility drills, flexibility and cardiovascular exercises. A few years later, Hejna et al (1982) studied high school athletes' injury rate and found that those who performed physical conditioning had far fewer injuries (26.2%) than their counterparts not engaged in physical conditioning (72.4%). Also, the time required for rehabilitation of the athletes who underwent the conditioning program was 2 days as opposed to 4.8 days for those who had not undergone the conditioning program. More recently, Hewett et al (1999) evaluated the effects of a 6-week conditioning program on the incidence of knee injury in high school female athletes and found a significantly lower incidence of knee ligament sprain or rupture than the control group. In the following year, Heidt et al (2000) reported that high school female soccer players who underwent pre-season conditioning had significantly fewer lower-limb injuries than those without the conditioning.

Notwithstanding the positive reports from the previous studies about the effects of physical conditioning on the musculoskeletal system, little information is available regarding the effect of physical conditioning on the patellofemoral joint and its surrounding soft tissues with different exercise protocols. Therefore, the present study was conducted to evaluate the effects of two weight training programs on the change in size of the VMO; mobility of the patella; patellar position; strength of the knee extensors; recruitment and onset timing of vasti muscles; as well as knee joint proprioception before and after the weight training programs. These outcome measures were selected since they were regarded as predisposing factors for PFPS (table 6-1).

Table 6-1: Measurable quantities related to PFPS

Quantity	*Author(s) and finding*	*Study design*	*Measurement*
Quadriceps thickness	Doxey (1987): Quadriceps thickness was significantly thinner in males with PFPS than the controls.	Retrospective, cross-sectional	B-mode ultrasound
VMO/VL amplitude ratio	Souza & Gross (1991): Significantly decreased VMO/VL ratio among subjects with PFPS compared to controls in isometric knee extension.	Retrospective, cross-sectional	Surface EMG
	Boucher et al (1992): Same as above.		
	Makhsous et al (2004): Same as above.		
Patellar tilt angle	Kujala et al (1993): Positive correlation between patellar tilt angle and knee pain score in patients with patellofemoral pain	Retrospective, cross-sectional	MRI
	Pookarnjanamorakot et al (1998): Increased patellar tilt angle in persons with anterior knee pain when compared with asymptomatic subjects.		

Table 6-1: Measurable quantities related to PFPS (*continued*)

Quantity	*Author(s) and finding*	*Study design*	*Measurement*
VMO-VL onset time difference	Cowan et al (2001): Subjects with PFPS' VL onset time occurred prior to that of VMO in stairs stepping, while the controls showed the opposite pattern.	Retrospective, cross-sectional	Surface EMG
	Boling et al (2006): Same as above. Also the author found PFPS subjects' EMG pattern was changeable after therapeutic exercises.	Prospective, longitudinal for 6 weeks	
Passive lateral patellar mobility	Witvrouw et al (2000): Increased lateral patellar mobility had a significant correlation with incidence of PFPS.	Prospective, longitudinal for 2 years	Custom-made measurement device
Knee joint proprioception	Baker et al (2002): Decreased knee joint position sense in persons with PFPS when compared with subjects without symptoms.	Retrospective, cross-sectional	Optical motion capture system
	Hazneci et al (2005): same as above.		Cybex dynamometer
Knee extensor force	Callaghan & Oldham (2004): Difference of the isokinetic knee extension torque between affected and unaffected legs of patients with PFPS was significantly higher than that of controls.	Retrospective, cross-sectional	Isokinetic machine

6.2. Objectives

This study aimed to assess the effectiveness of two modes of weight training (muscle hypertrophy versus muscular strength) on active and passive patellar stabilizers of previously untrained persons. The comparison of the outcomes of two different exercise protocols would have the clinical significance as a reference for athletic trainers and rehabilitation practitioners who prescribe exercises for athletes and patients.

6.3. Hypotheses to be tested

The hypotheses of this study were that weight training would:

1) enlarge the VMO cross-sectional area.

2) alter the patellar tilt angle.

3) alter the passive patellar mobility.

4) increase the knee extension torque.

5) alter the VMO-VL onset time difference.

6) alter the VMO/VL ratio of EMG amplitude.

7) alter the knee joint position sense.

6.3. Methods

Subjects

Fifty-three volunteers (37 males and 16 females) aged between 18 and 35 years (mean = 29.2, SD = 4.9) participated in this study. The recruitment criteria were that participants had:

1) never engaged in regular weight training;

2) no known pathology of lower extremity and torso;

3) no pain or injury of their lower limbs and torso for at least 6 months prior to the study;

4) no metal implant inside body (except permanent dental implant);

and were:

5) free from medical conditions including cardiovascular, respiratory, endocrine and metabolic diseases or any conditions that require periodical checkup or medication;

6) not pregnant.

The study was approved by the Human Ethics Sub-committee of the Hong Kong Polytechnic University (Appendix 5), and all subjects were requested to give written consent before their participation in the study (Appendices 6.1~6.4).

Five volunteers had withdrawn during the study; four subjects claimed training schedule conflict with their works, one declared that he lost his interest. The reminding forty-eight subjects completed all the required trainings and tests.

List of apparatus

1) Flexcomp computerized data acquisition system (Thought Technology, Montreal, Quebec, Canada) (fig 6-1).
2) Single-differential EMG electrode (Myoscan, Thought Technology, Montreal, Quebec, Canada) (fig 6-1).
3) Electrogoniometer (SG150, Biometrics, Cwmfelinfach, UK) (fig 6-1).
4) Disposable silver/silver chloride surface adhesive electrode (Multi Bio Sensors, El Paso, TX, USA).
5) Skin abrasive gel (NuPrep, DO Weaver & Co., Aurora, CO, USA).
6) Conductive gel (Signa Gel, Paker Laboratories Inc., Fairfield, NJ, USA).
7) Skin impedance meter (1089MKIII, UFI, Morro Bay, CA, USA).
8) Voltage isolator (T9405, Thought Technology, Montreal, Quebec, Canada).
9) Isometric dynamometer (Cybex Norm, Henley Healthcare, Nauppauge, NY, USA).
10) Medicine ball (2.7kg, Sissel, Inc., Sumas, WA, USA).
11) MRI scanner (Magnetom Avanto, 1.5 Tesla, Siemens AG, Erlangen, Germany).
12) B-mode ultrasound scanner with a 4 MHz annular array probe (Voyager, Ardent Sound, Mesa, AZ, USA).

13) Ultrasound gel pad (Aquaflex, Parker Lab, Fairfield, NJ, USA).

14) Hardness gauge (PTC Instruments, Los Angeles, CA, USA) (fig 6-8).

Modes of weight training

Although most previous studies (Berger 1962; Kaneko et al 1983; Kraemer et al 1995, 1997; Campos et al 2002; Ronnestad et al 2007) on weight training prescription for specific neuromuscular adaptations tended to adopt small sample sizes and short training periods, the majority of these studies concluded that if the primary goal of weight training is muscle hypertrophy while strength gain is secondary, the training protocol should be of moderate loading and repetition (e.g. 4 sets × 8-12 repetition maximum). Conversely, if strength enhancement is the principal purpose, then the training prescription would be of high loading and low repetition (e.g. 5 sets × 2-6 repetition maximum). These training principles have been accepted by the National Strength and Conditioning Association, USA (Baechle & Earle 2000).

Based on the above training principles, the subjects were quasi-randomly allocated into one of three groups, namely, muscle hypertrophy training (program H), muscle strength training (program S) and no-exercise control group. In order to have equal number of males and females in each group, the stratified allocation for gender was applied. This study was not a blinded study, thus the assessor was not blind to the group allocation.

Both programs H and S contained 24 sessions distributed evenly over a period of 8 weeks with 3 sessions per week (table 6-2); at least 48-hour rest was allowed between the sessions. All sessions were conducted by the same trainer (certified strength and conditioning specialist) to each subject on an individual basis.

Table 6-2: Two modes of progressive and supervised weight training programs

	Program H*	Program S*
Exercise	Parallel squat Knee extension (fig 6-2)	Parallel squat Knee extension (fig 6-2)
Set × rep	4 × 10 for each exercise	5 × 5 for each exercise
Rest between set	1 minute	2 minutes
Intensity	10 repetition maximum	5 repetition maximum

* = All weight training was isotonic, the concentric or eccentric phase was approximately 1 second. Also, standardized stretching for quadriceps, hamstrings, gluteus and calf muscles was requested before and after each training session (fig 6-3); the session lasted for 30-40 minutes for each subject.

Figure 6-1: Flexcomp data acquisition system. The system had four EMG electrodes and two electrogoniometers connected.

Parallel barbell squat with Smith machine (Icarian, Sun Valley, CA, USA)

Leg extension with resistance (Apollo 350, Tuff Stuff, Pomona, CA, USA)

Figure 6-2: Weight training used in current study.

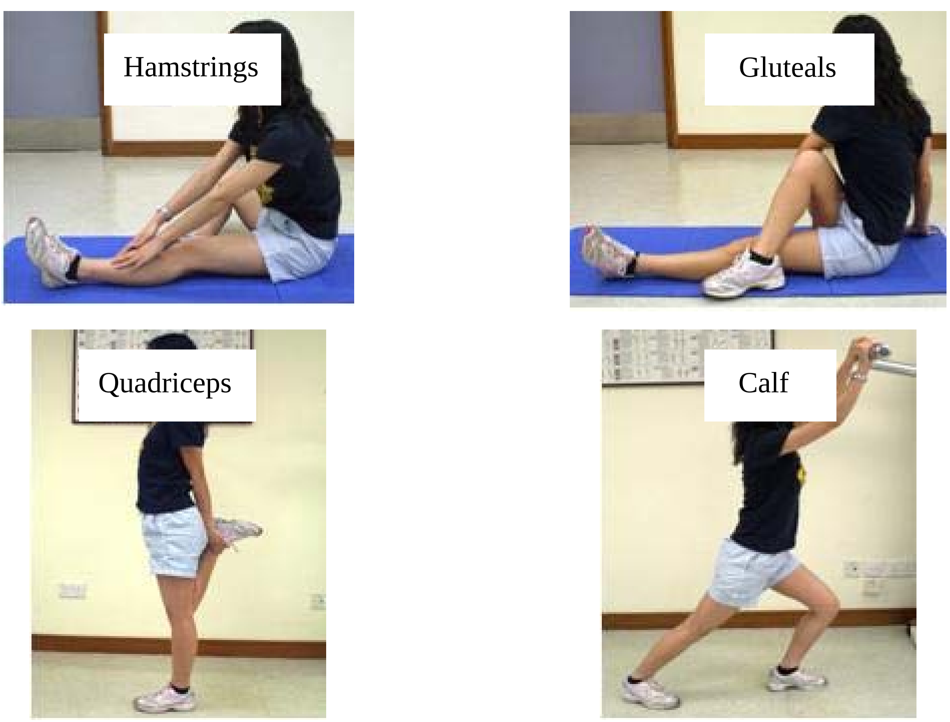


Figure 6-3: Standardized stretching. 10 seconds × 4 sets for each form.

Measurements

All the subjects were tested once before and once after the 8-week training, and had been asked to abstain from vigorous physical activities for 48 hours before the tests.

Cross-sectional area of VMO

Both MRI and B-mode ultrasound imaging techniques were used for measuring the cross-sectional area of VMO. Half of the subjects in each training group had MRI measurement taken before and after the 8-week training program. The other subjects in the training groups and all subjects in the control group took B-mode ultrasound imaging. The MRI (Beneke et al 1991; Mitsiopoulos et al 1998) and ultrasound imaging (Miyatani et al 2002; Wong & Ng 2007) have been confirmed to be valid and repeatable for measuring the vasti muscle cross-sectional area.

For the MRI examinations, subjects were supine, lying with knees extended and feet held together by adhesive tape. A 1.5 Tesla MRI scanner with surface coils (Magnetom Avanto, Siemens AG, Erlangen, Germany) was used to record the axial T1-weighted images (TR/TE: 420ms/50ms, slice thickness 3mm, inter-slice gap 0.3mm, field of view 35×20cm, matrix 448×256) of the subjects' knees and thighs. Foam blocks were used to support the surface coils in order to keep the anterior thighs and knees from being compressed by the coils. The MRI scanning lasted for 20-30 minutes and took place at the Queen Elizabeth Hospital, Kowloon, Hong Kong.

The image datasets were saved in DICOM format for off-line analysis. A line was drawn from the lateral knee joint line to the ipsilateral anterior superior iliac spine and the images obtained on a plane 10% proximal to the knee joint line were used to represent the VMO cross-sectional area (fig 6-4). The borders of the VMO muscle were traced manually with an image analysis software (MIPAV, version 2.0, National Institutes of Health, Bethesda, MA, USA) and a graphic digitizer (CTE-430, Wacom, Vancouver, WA, USA).

For the ultrasound imaging, the positioning of subject and identification of VMO followed the same procedures as the MRI. A diagnostic B-mode ultrasound scanner with a 4 MHz curved array probe (Voyager, Ardent Sound, Mesa, AZ, USA) was used to capture the VMO muscle cross-sectional area. The probe was totally supported by a custom-made mechanical stand that was adjustable in height and inclination angle (fig 6-5). A pre-molded ultrasound gel pad (Aquaflex, Parker Lab, Fairfield, NJ, USA) was used as an acoustic medium between the ultrasound probe and the VMO underneath. The ultrasound imaging lasted for 15-20 minutes and took place at the Department of Rehabilitation Sciences of the Hong Kong Polytechnic University.

The images captured for VMO were stored in DICOM format for later analysis with the Kodak 1D 3.5 software (Kodak, Rochester, NY, USA) by manual tracing with a graphic digitizer (CTE-430, Wacom, Vancouver, WA, USA). A pilot study was described in Chapter 5 of this thesis which revealed that the correlation between the MRI and the ultrasound imaging for the VMO muscle was significantly close (Pearson's coefficient = 0.97), and the between-day ultrasound measurement was repeatable (Intra-rater ICC (3,1) = 0.96).

Patellar tilt angle

Subjects in the training group who received the MRI also had their patellar tilt angle (PTA) measured at the mid-transverse patellar axial view (Shibanuma et al 2005). The PTA measurement was subtended by a line parallel with the lateral patellar facet and the posterior condylar line (fig 6-6). The measurement was reported to be reliable for between-day measurement, with an intra-rater ICC value of 0.95 (Sathe et al 2002).

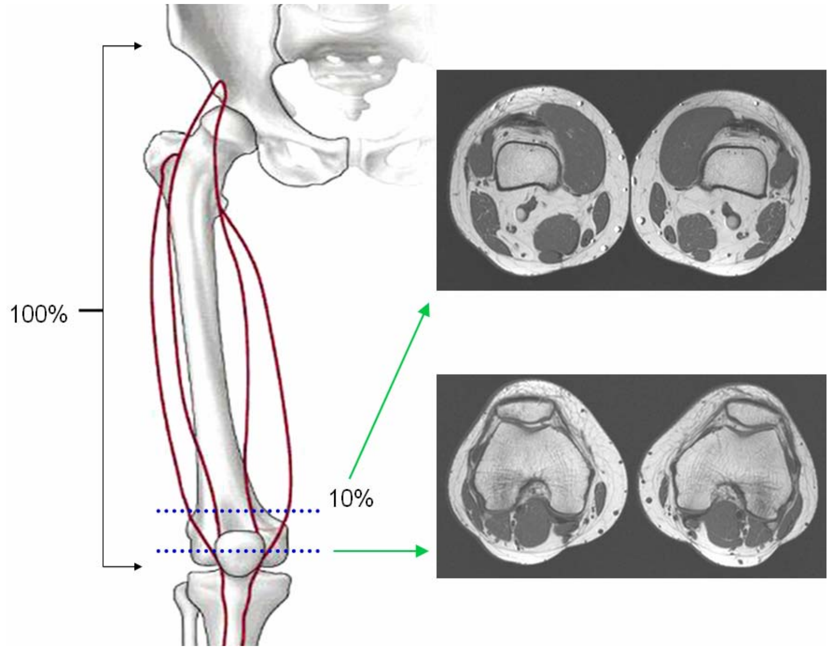


Figure 6-4: Selection of MRI images for analyzing VMO size and PTA. VMO muscle at distal 10% of the length between anterior superior iliac spine and lateral knee joint line, and the image at mid-transverse patellar level.

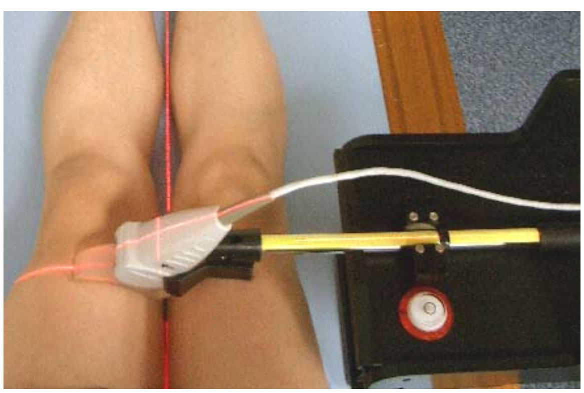


Figure 6-5: Hand-free ultrasound imaging for VMO cross-sectional area.

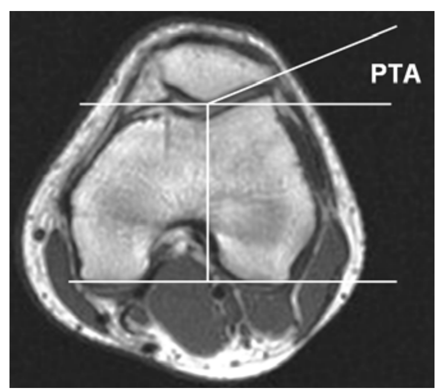


Figure 6-6: MRI based PTA measurement.

Passive lateral patellar mobility

The instrument developed to reliably measure the passive lateral patellar translation described in Chapter 4 was used in this study. Briefly, with the subjects in supine-lying, with knees extended and feet together, the ankles and lower legs were secured to restrict lower limb movements. A patellar glide apparatus (fig 6-7) consisting of an assembly with a force gauge (AEF-5, Aikon Engineering Co., Osaka, Japan), an electromagnetic motion tracker (pciBird, Ascension Technology Corp., Burlington, VT, USA), a custom made plastic linear slide and an adjustable platform (Poly-Jaque plastic lab jack, Scienceware, Pequannock, NJ, USA) was used to glide the subject's patella laterally at a constant velocity of 2 cm/s. The gliding force was 15 N, and the linear displacement of the patella was collected for analysis with Excel software (Office 2000, Microsoft, Redmond, WA, USA). Both knees were tested with three passive gliding tests and the average of these recordings was calculated to minimize random error.

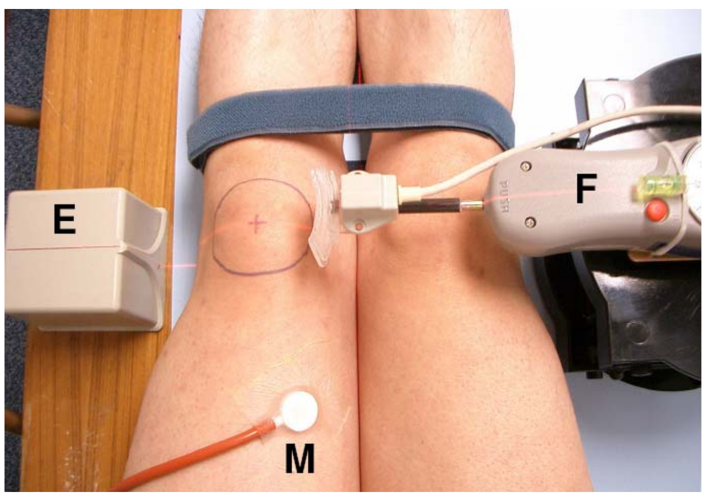


Figure 6-7: Setting of passive patellar mobility measurement. E = Electromagnetic tracker, M = MMG biofeedback, F = Force gauge

Isometric knee extension torque test

A dynamometer (Cybex Norm, Henley Healthcare, Nauppauge, NY, USA) was used to measure the slow-ramp maximal isometric knee extension torque of each subject with knee at 45° and hip at 85° of flexion. The detailed set-up followed the same procedure as the study described in Chapter 3. The test lasted for 5 seconds and was repeated three times with a 2-minute rest between trials. Only the highest force developed during the three trials was used for data analysis. The tests were performed twice at 8-week apart similar to all other measurements. The maximal isometric knee extension torque was reported as repeatable for between-day measurements with Pearson's coefficient of 0.88 (Welsch et al 1998), its within-day reliability measured by ICC was 0.93 (Bohannon 1990).

During the torque assessment, the peak EMG amplitude (root mean square, 125ms time constant) of VMO and VL corresponding to the highest torque were also recorded for normalizing the VMO/VL amplitude ratio in a functional movement described below.

The EMG amplitude ratio and onset time difference in VMO/VL during sit-to-stand

The set-up for surface EMG recording of VMO and VL followed the studies reported in Chapters 2 and 3. In brief, two single-differential EMG electrodes (Myoscan, Thought technology, Montreal, Quebec, Canada) were placed over the VMO and VL with disposable/adhesive circular snap buttons (Multi Bio Sensors, El Paso, TX, USA). The electrode for VMO was placed at 4 cm above the supero-medial corner of patella, whereas another electrode was placed over VL at 5 cm above the supero-lateral corner of patella (Cram & Kasman 1998; Wong & Ng 2006). The orientations of the VMO and VL electrodes were 55° and 15° to the longitudinal axis of femur respectively (Lieb & Perry, 1968).

A real-time data acquisition and display unit (Flexcomp system, Thought Technology, Montreal, Quebec, Canada) was used to record the EMG signals. The bandwidth of the signals was set at 20-500 Hz, notch filter was 50 Hz and the sampling rate was 1984 Hz. The raw data was later analyzed offline with Ergonomic Suite software (V1.52, Thought

Technology, Montreal, Quebec, Canada) for calculating the EMG onset and voltage amplitude of the VMO and VL.

An electrogoniometer (SG150, Biometrics, Cwmfelinfach, UK) was attached on the lateral right knee by double adhesive taping and connected to the Flexcomp system. Each subject sat on a 45-cm high chair without back support with feet at shoulder width apart, and the knees aligned to the vertical line from the toes. Three repetitions and comfortable rising pace were requested for each subject. On the verbal signal "go", the subject rose from this position to standing at a normal and comfortable pace.

The electrogoniometer recorded the dynamic knee joint angle and angular velocity (65ms moving-average). Only the data of EMG amplitude (root mean square, 125ms time constant) at 45° knee joint flexion was used to calculate the normalized VMO/VL amplitude ratio against the peak VMO and VL amplitude obtained from the maximal isometric knee extension torque test. The data of knee angular velocity at 45° of knee flexion was used to check the consistency of the sit-to-stand movements. A ± 5% of the velocity was regarded acceptable in repeated measurements of the pre/post training.

The EMG onset point of the VMO and VL was determined as at double of the baseline peak-to-peak amplitude, as described in Chapter 3. The VMO-VL onset time difference was calculated by subtracting the onset of VMO from the onset of VL. In order to minimize the random errors, the average of the VMO-VL onset difference and the normalized VMO/VL amplitude ratio of the three sit-to-stand motions were calculated for each subject.

A pilot study including five healthy subjects (10 legs from 3 males and 2 females) was done for test-retest reliability with a 7-day interval for the VMO-VL onset difference and the VMO-VL amplitude ratio in the sit-to-stand, the intra-rater ICC (3,1) (consistency) was 0.78 and 0.81, respectively.

VMO-VL onset time difference in knee perturbation

In this measurement, the participants were subject to knee perturbation at single-leg standing so as to induce a stretch reflex in the quadriceps muscles (Ng 2005; Fong & Ng 2006). With their ears shielded, subjects assumed alternative single-legged standing on both sides with the knee extended and the hands lightly touching the wall in front for balance. A medicine ball of 2.7 kg was used to provide perturbation to the knee joint. The ball was suspended with a sling from the ceiling of the room so that it could be adjusted to be at the level of the knee joint of the subjects. The researcher pulled the ball back until the sling made an angle of 45^o to the vertical behind the subject and upon release of the ball, it would exhibit a pendulum swing and hit the back of the subject's knee to produce a perturbation.

The perturbation test was repeated three times at 20-second intervals. The EMG onset point of VMO and VL was determined once the EMG amplitude was double the baseline EMG data (Wong & Ng 2005). In order to minimize random error, the average VMO-VL onset difference of the three knee perturbation trials was calculated for each subject.

A pilot study was performed with nine subjects (7 males and 2 females) for the same-day reliability of the knee perturbation test, the reliability (ICC (3,1), consistency) was 0.86. Ng (2005) reported that the between-day reliability of the test was 0.70 (ICC).

Knee joint position sense

The subject was seated on a couch with knees flexed and torso unsupported and the knees positioned over the edge of the couch 3 cm proximal to the popliteal fossa. An electrogoniometer (SG150, Biometrics, Cwmfelinfach, UK) was attached to the lateral knee joint and connected to the Flexcomp data acquisition system that displayed knee joint angle in real-time. An inflatable air splint (Model 70-008, Svend Andersen, Haarlev, Denmark) was applied to the lower leg and ankle throughout the test for neutralizing skin sensation. With the subjects' eyes closed to eliminate visual cue, the researcher moved the lower leg passively from the resting position (approximately 90° knee flexion) to pre-set positions of 30°, 40°, 50°, 60° and 70° in random orders. The subject was asked to hold the leg in the test position for 4 seconds while the leg was un-supported. Then the researcher brought the leg to its resting position. The subject was required to reproduce the previous test position and to hold at that position for 4 seconds.

Five measurements were taken for each knee. The mean difference between the tested angle and the subjects' reproduced angle was scored as the knee joint position sense test (Barrack et al 1983, Baker et al 2002) (fig 6-8). A pilot study including five healthy subjects (10 knees from 3 males and 2 female) was done for test-retest reliability with a 7-day interval for the knee joint position sense. The intra-rater ICC (3,1) (consistency) obtained from the pilot study was 0.88. Also, Petrella et al (1997) reported the between-day reliability of the electrogoniometic measurements for knee joint position test was high, as the Pearson's coefficient of the scores was $r = 0.89$.

The above tests included patellar mobility, EMG of vasti muscles, isometric force and knee joint sensation; the whole procedures lasted for 90-120 minutes and took place at the

Department of Rehabilitation Sciences of the Hong Kong Polytechnic University. For all the dependent variables, the testing order was as follows, MRI or ultrasound imaging, patellar mobility, knee joint position sense, EMG of vasti muscles and isometric force.

Statistical analysis

With the level of significance was set at 0.05, data analyses were performed using Statistical Analysis in Social Science software (V 11.5, SPSS Inc., Chicago, IL, USA) for:

1) Baseline characteristics: Between-group comparison by one-way ANOVA (table 6-3).

2)
- VMO cross-sectional area,
- Passive lateral patellar mobility,
- Knee extensor torque,
- VMO/VL amplitude ratio in sit-to-stand,
- VMO-VL onset time difference in sit-to-stand,
- VMO-VL onset time difference in knee perturbation,
- Knee joint position sense

⟹ Between-group comparison for pre- and post-training measurements with 3 × 2 ANOVA (defined as 3 groups × 2 measurements) (table 6-4). If any factor is significant, one-way ANOVA with Bonferroni adjustment was used to identify the pair of data that contributed to the significance (table 6-5). Within-group comparison with paired t-test was used to compare all three groups between the pre- and post-training for the each dependent variable (figs 6-11 ~ 6-16).

3) Patellar tilt angle: Pre- and post- training PTA was compared with 2 × 2 ANOVA (defined as 2 groups × 2 tests) (table 6-6).

Figure 6-8: Evaluation of knee joint position sense (above left). In order to standardize the pressure inside the air splint for the repeated measurements, a hardness gauge (PTC Instruments, Los Angeles, CA, USA) was used to monitor the surface hardness of the splint before the knee joint sense test (above right).

6.4. Results

1) Baselines

The three groups had comparable baseline characteristics (table 6-3) and pre-training measurement values (table 6-5).

Table 6-3: Subjects' baseline characteristics (mean ± SD).

	Program H (12 males, 4 females)	Program S (12 males, 4 females)	Control (12 males, 4 females)	p
Age	27.6 ± 5	26.9 ± 4.4	27.2 ± 5.5	0.93
Weight (kg)	63.1 ± 9.1	61.1 ± 8.1	63.4 ± 8.3	0.71
Height (cm)	168.3 ± 4.5	168.2 ± 5.8	170.2 ± 4.8	0.47
BMI	22.6 ± 2.9	21.5 ± 1.8	21.8 ± 1.9	0.34

2) Between-group comparison by 3×2 ANOVA

Table 6-4: Overall between-group comparison.

Parameter	Group	Test	Group×test
	P value		
VMO size	<0.001*	0.007*	0.18
Patellar mobility	0.259	<0.001*	0.029*
Knee extension force	<0.001*	<0.001*	<0.001*
VMO/VL EMG ratio	0.278	0.012*	0.28
VMO:VL onset difference in sit-to-stand	0.944	0.092	0.572
VMO:VL onset difference in knee perturbation	2.246	0.307	0.068
Knee joint position sense	<0.001*	<0.001*	<0.001*

* = Statistical significance

VMO cross-sectional area

Between-group comparison: There was no significant difference between the groups before the training program (p>0.05). After training, the size of VMO in both training groups was significantly different from that of the controls (p=0.01) (table 6-5), but the post-hoc analysis showed that there was no significant difference between these two training groups in terms of the muscle size gain (p>0.05).

Within-group comparison: After the training, VMO size in the H and S training groups enlarged significantly (p<0.001 for both H and S) (fig 6-9, 6-10), while the change in the control group was not significant (p>0.05) (fig 6-11).

Passive lateral patellar mobility:

Between-group comparison: There was no significant difference between the groups before the intervention ($p>0.05$). After the training, the patellar mobility of both training groups was significantly different from that of the controls ($p=0.01$) (table 6-5). Post-hoc analysis revealed no significant difference between these two groups in terms of the change of passive lateral patellar range of motion ($p>0.05$).

Within-group comparison: The patellar mobility decreased significantly in both training groups after the training program ($p<0.001$ for both H and S), while no significant change was observed for the control group ($p>0.05$) (fig 6-12).

Knee extension torque:

Between-group comparison: There was no significant difference between the groups before the program ($p>0.05$). After the training, the torque of both training groups was significantly different from that of the controls ($p<0.001$) (table 6-5). Post-hoc analysis revealed no significant difference between these two groups in terms of increase in quadriceps strength ($p>0.05$).

Within-group comparison: All three groups showed the torque increased significantly at the post-training measurement ($p<0.001$ for all three groups), but the intervention groups had more than 45 Nm increment in average, while the control group increased less than 3 Nm. (fig 6-13).

VMO/VL amplitude ratio in sit-to-stand:

Between-group comparison: There was no significant difference between the groups before and after the 8 weeks ($p>0.05$) (table 6-5).

Within-group comparison: The amplitude ratio raised significantly in both training groups after the 8-week exercise ($p<0.001$ for H and S), while the control group remained statistically unchanged ($p=0.63$) (fig 6-14).

VMO:VL onset time difference in sit-to-stand:

Between-group comparison: There was no significant difference between the groups before and after the 8 weeks ($p>0.05$) (table 6-5).

Within-group comparison: The VMO:VL onset time difference changed significantly in both training groups after the 8-week exercise ($p=0.11$ for H; $p=0.07$ for S), while the control group remained statistically unchanged ($p=0.47$) (fig 6-15).

VMO-VL onset time difference in knee perturbation:

Between-group / within-group comparison: In the knee perturbation test, all three groups showed no significant difference at both pre/post-training tests ($p>0.05$) (table 6-5).

Knee joint position sense:

Between-group comparison: There was no significant difference between the groups before the training ($p>0.05$). After the training, the position sense of both training groups was significantly different from that of the controls ($p<0.001$) (table 6-4). Post-hoc analysis revealed no significant difference between these two groups in terms of the improvement on knee joint position sense ($p>0.05$).

Within-group comparison: The position sense changed significantly in both training groups after the 8-week exercise ($p<0.001$ for both groups), while the control group remained statistically unchanged ($p>0.05$) (fig 6-16).

3) Patellar tilt angle

For the PTA, there was no significant difference between the subjects in program H and S, also the training did not significantly change the PTA in both groups (table 6-6).

Table 6-5: Pre- and post- training measurements among groups by one-way ANOVA. Values showed in mean ± SD. * = statistical significance

		Program H (n=16)	Program S (n=16)	Control (n=16)	F	p
VMO size (cm^2)	Pre	11 ± 3.4	12 ± 2.3	10.5 ± 2.2	2.43	0.09
	Post	12.8 ± 4.6	13.6 ± 2.7	10.6 ± 2.3	7.32	0.01*
Lateral patellar mobility (mm)	Pre	14.3 ± 3.2	14.2 ± 3.6	13.7 ± 2.9	0.26	0.77
	Post	11.7 ± 2.9	11.5 ± 2.7	13.6 ± 3	5.39	0.006*
Knee extensor torque (Nm)	Pre	90.6 ± 22	94.1 ± 24.5	92.3 ± 20.1	0.18	0.84
	Post	135.7 ± 36.6	139.3 ± 29.7	94.9 ± 20.5	22.1	<0.001*
VMO/VL amplitude ratio in sit-to-stand	Pre	0.89 ± 0.23	0.91± 0.21	0.87± 0.19	1.85	0.16
	Post	1.13 ± 0.24	1.14 ± 0.3	0.88 ± 0.19	0.67	0.51
VMO-VL onset time difference in sit-to-stand (ms)	Pre	-3.2 ± 12.6	-4.2 ±11.2	-1.9 ± 11.5	0.29	0.75
	Post	0.5 ± 11.3	0.1 ± 10.1	-1.5 ± 10.8	0.33	0.72
VMO-VL onset time difference in knee perturbation (ms)	Pre	2.2 ± 4.8	1.0 ± 4.9	2.5 ± 4.2	1.01	0.36
	Post	2.8 ± 3.9	1.3 ± 3.8	2.6 ± 4.0	1.36	0.26
Knee joint sense (degree)	Pre	3.0 ± 0.7	3.1 ± 0.9	3.0 ± 0.8	0.33	0.72
	Post	1.7 ± 0.3	1.8 ± 0.3	2.9 ± 0.7	79.95	<0.001*

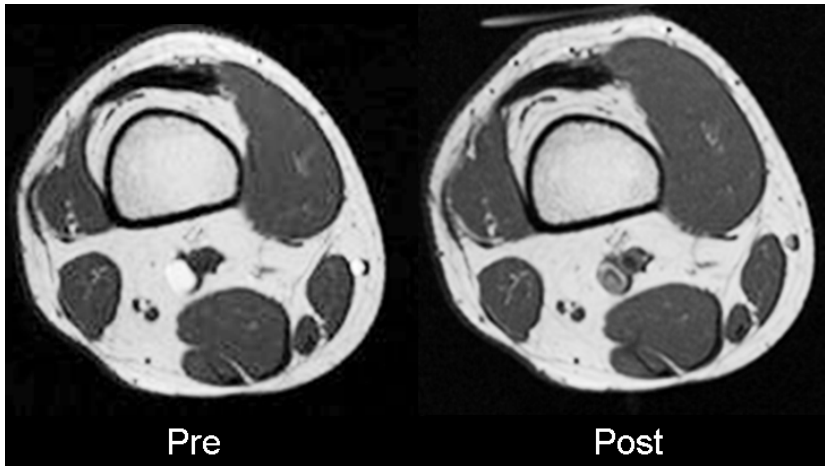


Figure 6-9: Pre and post-training MR image of the distal thigh at suprapatellar level. Images were selected from a subject of program H.

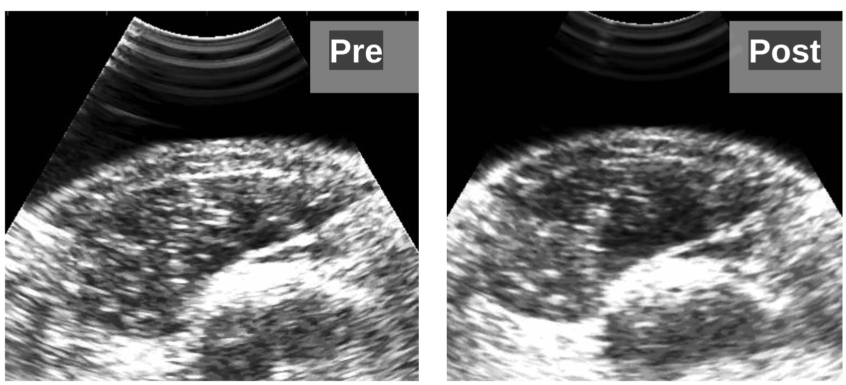


Figure 6-10: Pre and post-training ultrasonic image of the VMO. Images were selected from a subject of program S.

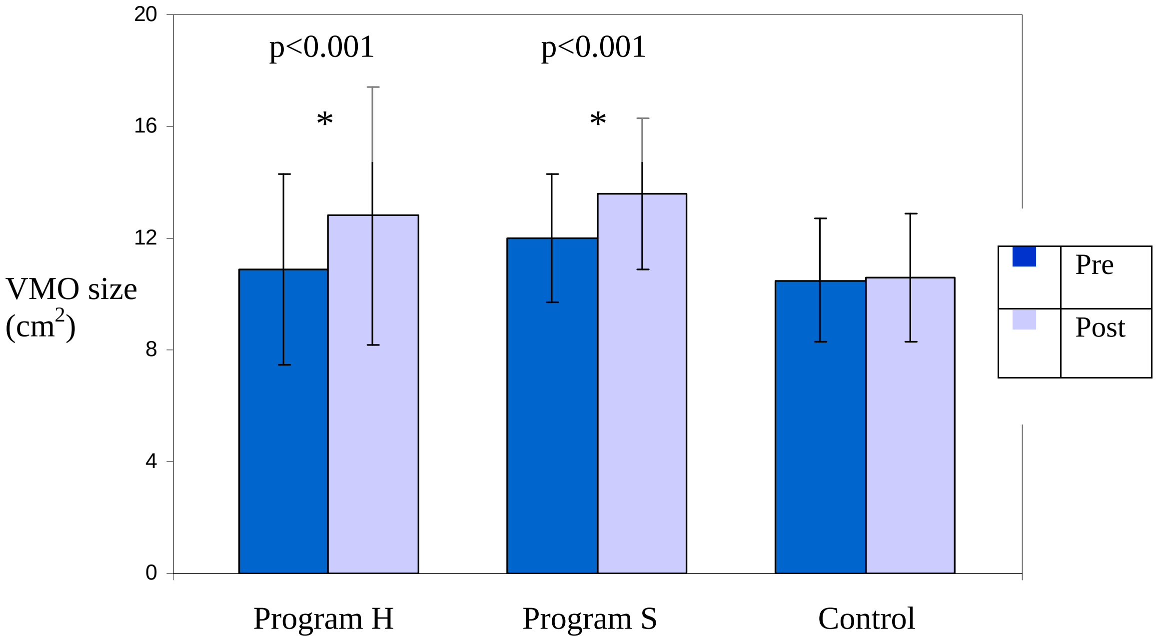


Figure 6-11: Comparison for pre- and post-training VMO size by paired t-test. Bar chart shows mean ± SD, * = statistical significance

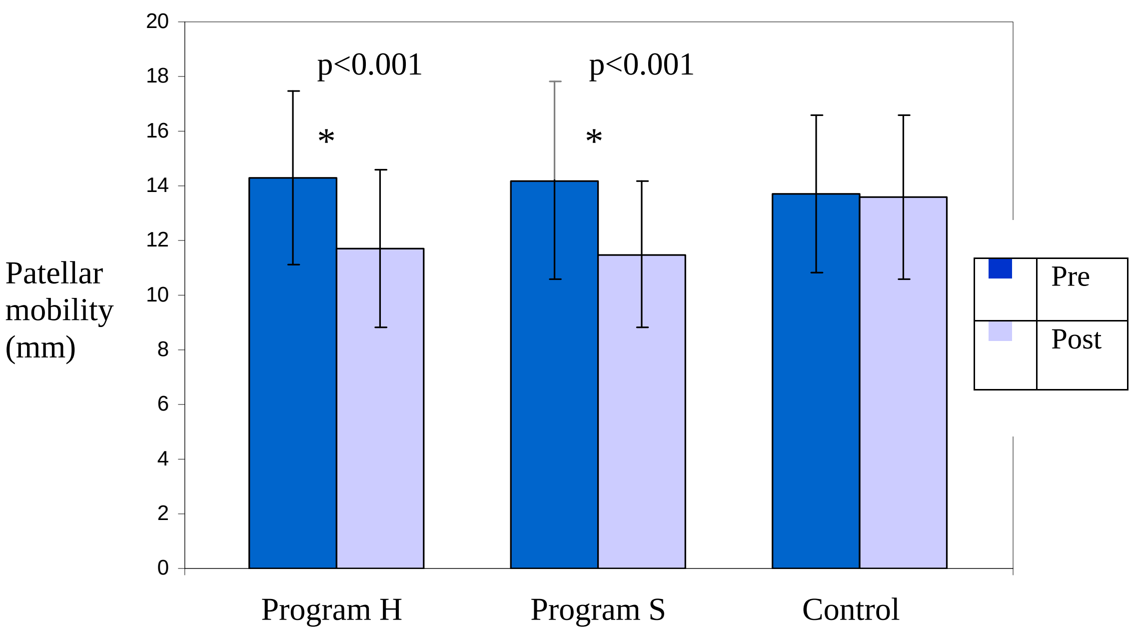


Figure 6-12: Comparison for pre- and post-training lateral passive patellar mobility by paired t-test. Bar chart shows mean ± SD, * = statistical significance

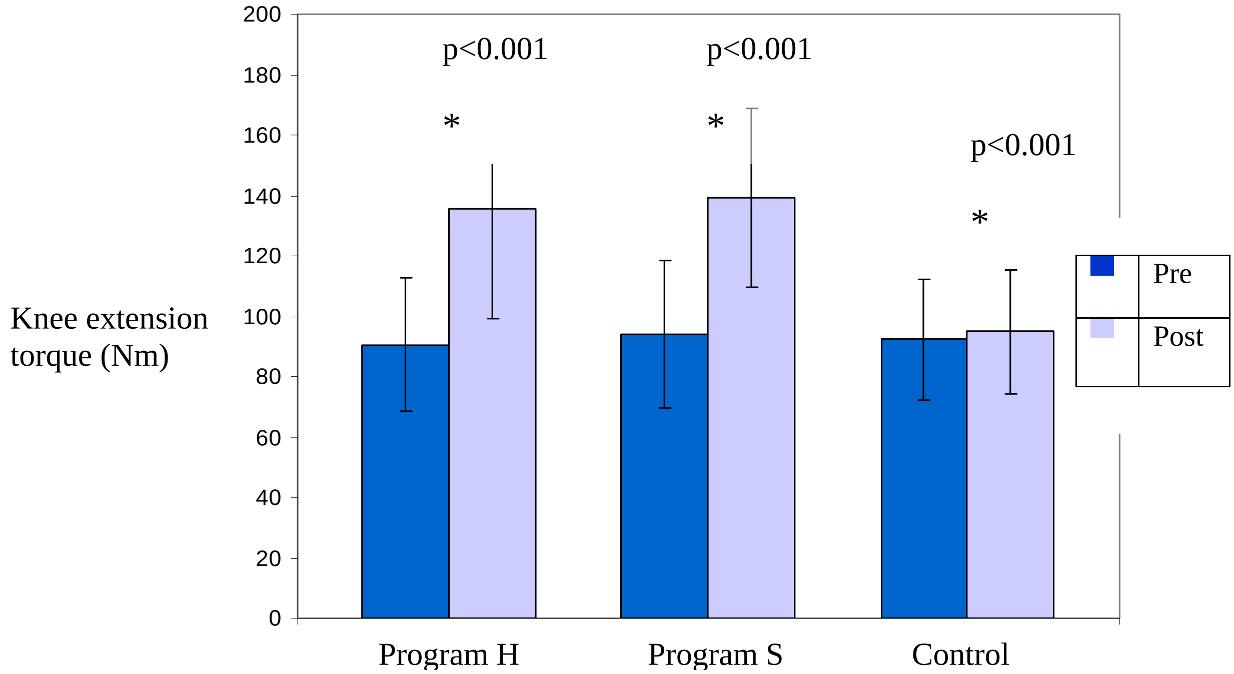


Figure 6-13: Comparison for pre- and post-training knee extension torque by paired t-test. Bar chart shows mean ± SD, * = statistical significance

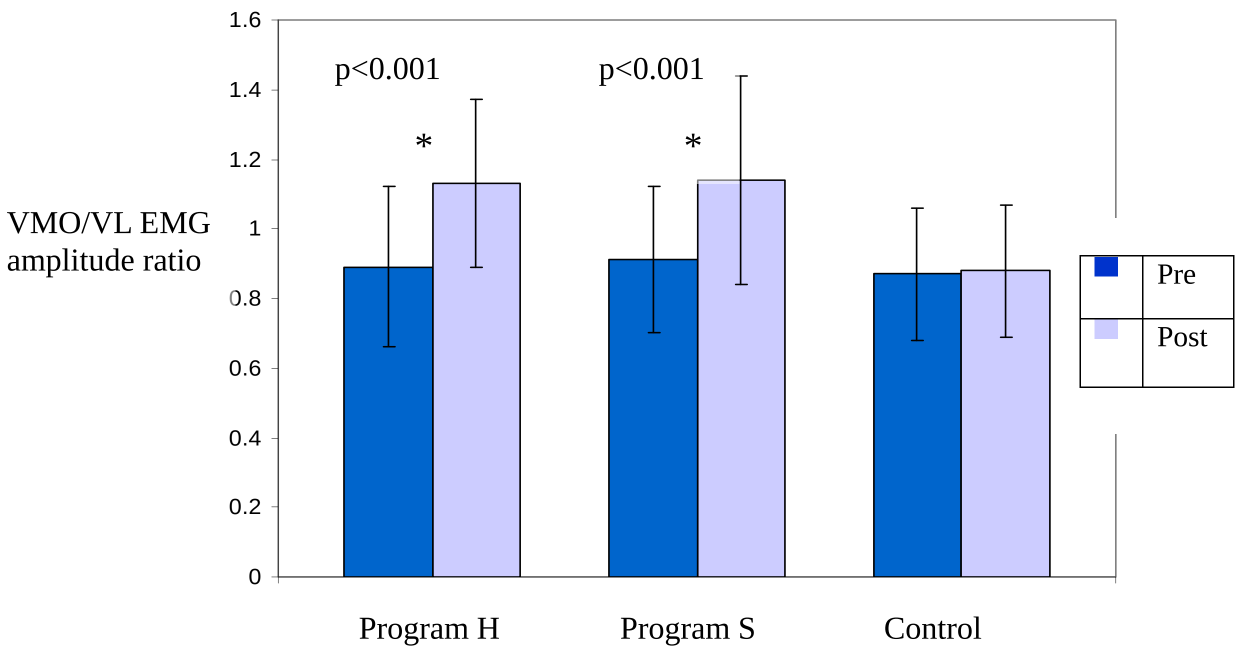


Figure 6-14: Comparison for pre- and post-training VMO/VL amplitude ratio in sit-to stand by paired t-test. Bar chart shows mean ± SD, * = statistical significance

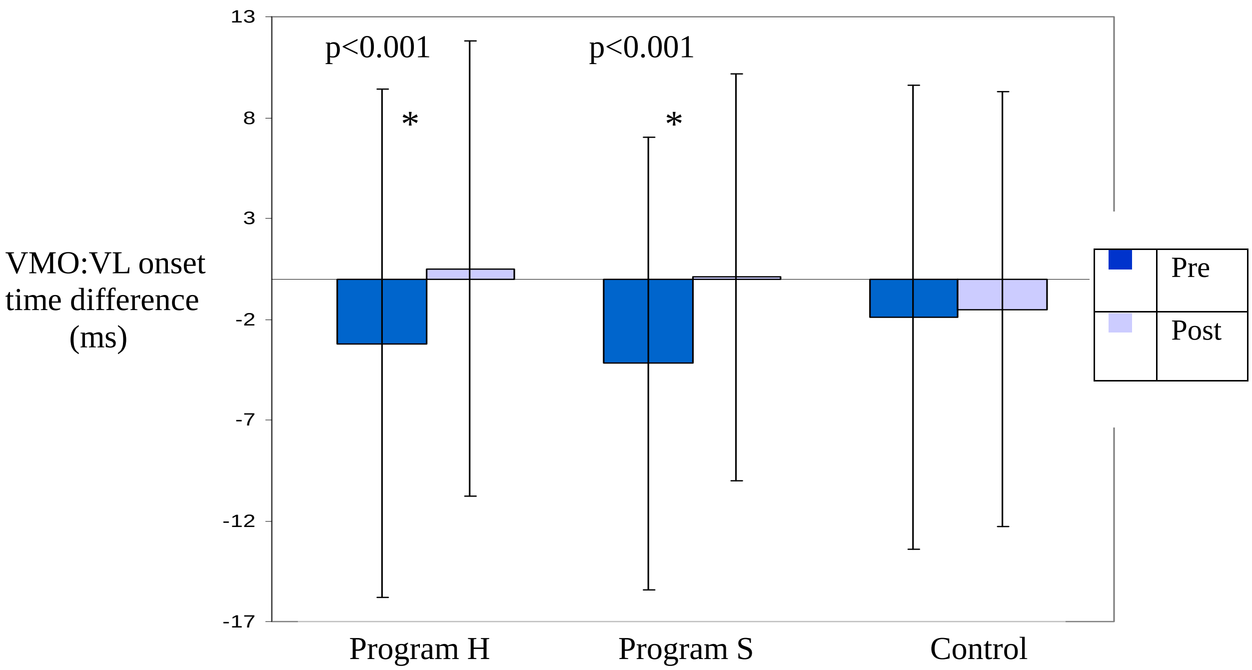


Figure 6-15: Comparison for pre- and post-training VMO:VL onset time difference in sit-to-stand by paired t-test. Bar chart shows mean ± SD, * = significance

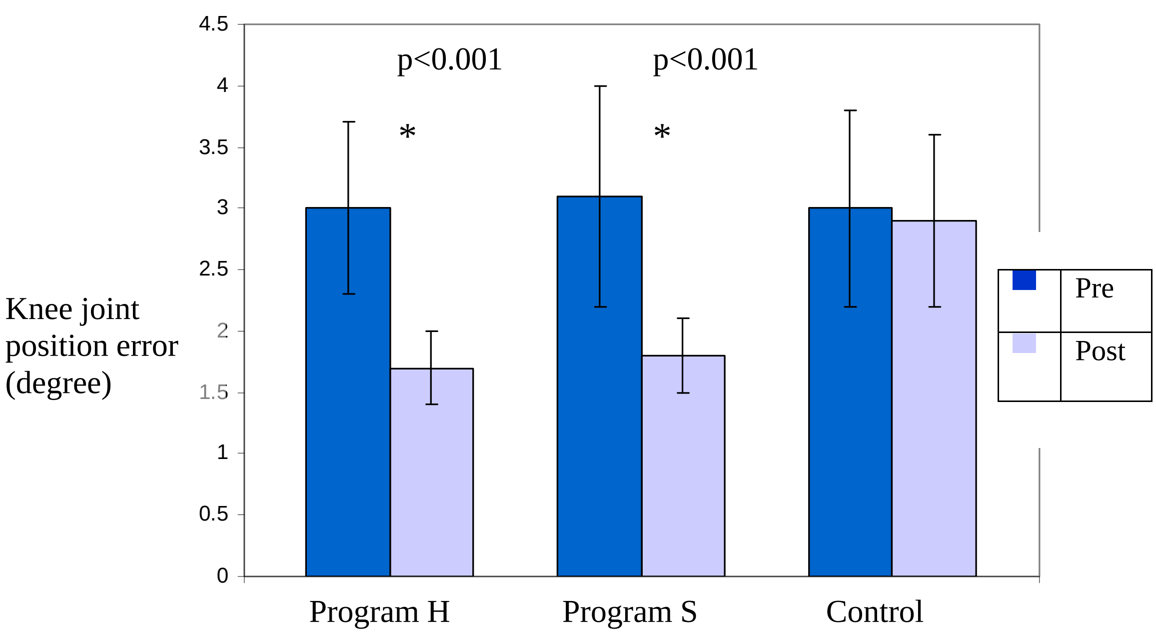


Figure 6-16: Comparison for pre- and post-training knee joint position error by paired t-test. Bar chart shows mean ± SD, * = statistical significance

Table 6-6: Comparison for pre- and post-training PTA values by 2 × 2 ANOVA

Values showed in mean ± SD, * = statistical significance

		Program H (n=8)	Program S (n=8)	Control	p (group)	p (test)
Patellar tilt angle (degree)	Pre	14.6 ± 4.7	12.8 ± 4.5	not available	0.12	0.87
	Post	14.8 ± 3.5	13 ± 5.1			

6.5. Discussion

This study demonstrated that 8 weeks of either weight training program targeted to improve muscle strength or muscle size resulted in comparable changes in knee joint position sense, surface EMG characteristics, quadriceps strength, passive patellar stability, and hypertrophy of the VMO muscle in previously untrained subjects.

A weak VMO has been reported to be a contributing factor of PFPS (Fox 1975; Fisher 1986; Wilk et al 1998). Since muscle strength and size are positively correlated, although the correlation is not necessarily linear (Davies 1990; Fukunaga et al 2001), a larger VMO muscle after training would imply a potentially greater medial stabilizing force on the patella. Our present findings showed that both modes of weight training being effective for VMO muscle building have vital clinical implications in the prevention of PFPS in light of the strategic importance of VMO in maintaining normal patellar tracking (Neptune et al 2000; Cowan et al 2001; Ng et al 2006), and weakness of this muscle is considered to be a causative factor of PFPS (Souza & Gross 1991; Boucher et al 1992; Makhsous et al 2004).

Hyper-mobility of patella has been identified as a predisposing factor for PFPS in both cross-sectional and prospective studies (Kujala et al 1986; Witvrouw et al 2000). The lateral patellar mobility measurement indicated that both weight training programs could enhance the patellar stability on the femoral trochlea. The medial patellofemoral ligament (MPFL) that bridges the adductor tubercle and the medial patella, constitutes 53% of the resistance to the lateral patellar translation at full knee extension, while the medial patellomeniscal ligament (MPML) constitutes 22% of the resistance (Colan et al 1993; Hautamaa et al 1998). The present finding of a decreased lateral patellar translation after weight training is suggestive of a strengthening effect on MPFL and/or MPML as a result of biological tissue adaptation. However, since the lateral patellar mobility test was performed in a quasi-static manner, the effect of weight training on the dynamic patellar tracking during faster and more functional movements warrants further investigation.

The VMO/VL EMG amplitude ratio of VMO/VL in isometric knee extension has been found to be different between asymptomatic subjects and individuals with anterior knee pain (Souza & Gross 1991; Boucher et al 1992; Makhsous et al 2004), the latter had relatively lower VMO/VL activity ratio. This distinguishable EMG ratio is regarded as a consequence of the VMO contributing less to the total knee extension torque and the imbalanced load sharing between VMO and VL, thus directing the patella towards the lateral side (Makhsous et al 2004). The present study found that subjects in the training groups had increased VMO/VL activity ratio in sit-to-stand after the training program, signifying greater recruitment of VMO muscle (Conwit et al 1999), enhanced VMO muscle contractile force (Suzuki et al 2002) and greater contribution of the VMO in the functional movement.

The VMO:VL onset time difference in sit-to-stand can provide a window for observing the motor control of the VMO and VL as synergists. The sit-to-stand action is a pre-programmed movement that does not rely heavily on conscious sensory feedback (Lephart et al 1998a). The present study showed this feedforward or open-loop motor control of the VMO and VL was significantly altered by the weight trainings. The pre-post training difference was approximately 4 ms in VMO onset after the training in both intervention groups. But it is not known if the 4-ms change is sufficient to make a difference to the patellar tracking or patellofemoral joint contact area (Voight & Weider 1991; Grabiner et al 1994; Powers et al 2004; Ota et al 2006).

For VMO:VL onset time difference, subjects demonstrated more variability in sit-to-stand than in knee perturbation. The VMO:VL onset time difference in knee perturbation was not statistically different in both between-group and within-group comparisons. The knee perturbation itself is similar to the knee jerk reflex test. The quick stretch induced by the perturbed knee activated muscle spindles in the quadriceps, the type Ia sensory axons from the spindles carry action potentials to the spinal cord where they synapse directly on motor neurons of the same muscle that was stretched, thus a brief and involuntary quadriceps contraction is induced (Toft et al 1989). The knee jerk reflex is a neonatal reflex present at birth. Although medication (Pickar 1998), muscular fatigue (Lam et al 2002), and physical exercise (Rittweger et al 2003) can have temporary effects on either enhancing or inhibiting neuromuscular excitability of the reflex, the present study did not find evidence that VMO:VL onset time difference in knee perturbation can be altered by the 8-week weight training program.

In order to prevent non-contact knee injuries during running, cutting or landing, sensitive and accurate knee joint proprioception is required (Lephart et al 1998b; Williams et al 2001). Two studies have reported that persons with PFPS had relatively less accurate knee joint position sense as compared with the controls (Baker et al 2002; Hazneci et al 2005). Since the studies were retrospective, they could not determine if the defective knee joint position sense was a contributing factor or an effect of the PFPS. In addition, Jerosch & Prymka (1996) reported that the knee joint position sense in patients with recurrent patellar dislocation was reduced compared to that in the controls. The diminished knee joint position senses were revealed in both the affected and the unaffected knee compared with the controls. As the present study showed the joint sense could be improved significantly after the 8-week training. The enhanced knee joint sensation by regular weight training may provide a preventive effect against non-contact incidence of patellofemoral disorders (Bernauer et al 1994; Kibler et al 2001).

Ingersoll and Knight (1991) reported that asymptomatic subjects undergoing only 3 weeks of resistance or EMG biofeedback training to the quadriceps were able to change their patellofemoral joint congruence under the X-ray Merchant's view examination. They hypothesized that the shortened knee extensor muscles could induce a more favorable patellar tracking, thus reducing the predisposition to lateral patellar instability. In the present study, the Merchant's method was not used since it is less reliable than the PTA measurement and also not sensitive enough in detecting small changes of patellar tilting (Inoue et al 1988; Delgado-Martinez et al 2000). It has been reported that PTA of less than 8° is closely correlated with the presence of PFPS (Schutzer et al 1986; Pookarnjanamorakot et al 1998). Among the 16 subjects included in the PTA examination, only two subjects had a marginally abnormal PTA (6 - 7.5°). Therefore, the finding of no significant change in PTA after the training may be due to a ceiling effect, as most of the subjects had normal PTA before the study.

Previous studies regarding weight training prescriptions (Berger 1962; Kaneko et al 1983; Kraemer et al 1995, 1997; Campos et al 2002; Ronnestad et al 2007) supported the concept that different combinations of repetition, set and intensity would produce different degrees of muscular strength and hypertrophy enhancement. Physiologically, only motor units that are recruited in the exercise are subject to adaptational changes with weight training. A motor unit is composed of either all slow-twitch or all fast-twitch muscle fibers. According to Henneman's size principle for recruitment of the motor units, the low-threshold motor units that are composed of slow-twitch muscle fibers are recruited first. As the demands of weight training rising with higher resistance, progressively higher-threshold motor unit that are composed predominantly of fast-twitch fibers would be recruited (Henneman et al 1974; Ebbeling & Clarkson, 1989; Duchateau et al 2006). Thus, the weight training for muscular strength (program S) could recruit more muscle fibers than the training for muscle hypertrophy (program H). While the program H did provide longer and more frequent stimulus for the muscle fibers in a given session than the program S, since a relatively higher number of repetitions and shorter rest times are required for the program H.

The present findings indicated that both weight training programs, H and S, led to comparable gain on VMO muscle mass and knee extension torque, although their training prescriptions were diverse in numbers of repetition and set, duration of rest between the sets, and loading (table 6-2). It is believed that the length of 8-week weight training which consisted of 24 sessions might not be long enough to initiate the detectable difference in the outcome measurements between the program H and S for previously untrained persons.

The subjects were requested to maintain their life styles including habit of diet and pattern of sleep during the 8 weeks. It is unknown if these confounding factors could affect the outcomes, because as Biolo et al (1997) and Esmarck et al (2001) reported, skeletal muscle building is a nutritionally modulated process, supplementary amount of protein intake can enhance the muscle synthesis after exercise. Also, inadequate sleep can hinder the weight training performance and exercise induced muscle growth (Reilly & Piercy 1994).

During the 8-week training programs, the compliance of subjects was reasonably satisfactory, 28 out of 32 intervention group subjects completed all 24 sessions while 4 subjects completed 21 sessions. Also, no physical injury was reported during the 8 weeks and five intervention group subjects quitted due to personal reasons.

In light of the high prevalence of PFPS among athletes, an effective prophylactic training program could have a significant impact on the athletes' career. This study is believed to be the first of its kind to examine the effects of different modes of weight training on the quadriceps strength, size of VMO muscle and patellar stability. The positive findings suggest that a prescribed and supervised weight training program could improve the patellar stabilizers and patellar stability which may be crucial in the prevention of PFPS for sedentary people and individuals engaged in sports activities.

6.7. Conclusion

Both weight training programs lasting for 8 weeks and that are targeted to build muscle mass or muscle strength could lead to significant hypertrophy of the VMO, increased passive lateral patellar stability, enhanced knee extension force, quicker VMO onset, greater VMO/VL amplitude ratio and more accurate knee joint position sense for previously untrained adults as compared with the controls. There was, however, no discernible difference between the two weight training programs on the parameters tested.

Chapter 7

Grand Discussion and Conclusion

The main objectives of this thesis are to investigate the effect of weight training on the biomechanical, neurological and morphological adaptations of the active and passive patellar stabilizers. In addition, the different outcomes induced by two modes of weight training are evaluated and compared.

The Chapters 2 to 5 were individual experiments in which different biomechanical and imaging instruments or methods developed for this thesis were evaluated and validated. The instruments and methods served as building blocks of the main study (Chapter 6).

7.1. Weight training

The weight training programs utilized in Chapter 6 were in form of a continuous, non-explosive, non-assistive, and repetitive pattern. In performing the parallel barbell squat, conscious control of muscle contractions and posture, awareness of dynamic balance and joint positions are required. The squat exercise, which involves hip, knee and ankle joint motions, was entirely performed by the subject without additional visual cues or feedback from a mirror. In the beginning sessions, verbal feedback was provided by the researcher and close monitoring was kept throughout the 8-week training. The verbal feedback was not real time because when the researcher observed an imperfect form of the subjects' exercise, such as over-/under-depth of knee bending or excessive forward knee position, the researcher would tell and teach the subject the proper form, so that the subject could correct the form in the next repetition (Wong 1994). In light of the fact that the skillfully precise movement of parallel squat would largely rely on the subject's own motor control; in other words, muscle synergism and co-contraction, propriception awareness and pacing of dynamic movement were called for the nature of the squat exercise. In contrast, the knee extension exercise requires only knee joint movement and may demand relatively less on the lower limb proprioception since the subjects can actually see their lower legs moving even without a mirror. Yet the voluntary control of the knee angular velocity is still critical in executing the concentric and eccentric phase of the knee extension exercise. Based on the demands of the exercises, the practice of 8-week weight training, either for muscle hypertrophy or strength, is beneficial for the physiological adaptations in terms of knee joint proprioception, knee extensor force development, quadriceps muscle recruitment and growth. These have been confirmed by the outcome measures in the present study.

In Chapter 6, both muscle hypertrophy and strength training were found to be equally effective for improving the contributing factor or measurable quantities related to the patellofemoral pain syndrome (PFPS). It is not known if a longer training program that lasted for more than 8 weeks would lead to a different outcome. It is particularly questionable if the muscle strength program would produce relatively greater knee extension torque and the muscle hypertrophy program would build greater VMO muscle mass (Fry 2004). Various studies had reported that the exercise prescription for effective muscular strength gain varies from 2 RM (repetition maximum) to 10 RM with multiple sets (Berger 1962; Kaneko et al 1983; Kraemer et al 1995, 1997; Campos et al 2002; Ronnestad et al 2007), and these are consistent with the present study that compared the effectiveness of the 5 RM and 10 RM, and the present result revealed that the muscular strength improvement between the 5 RM × 5 sets versus 10 RM × 4 sets was not significant different after the 8-week training.

This study shows that resistance training (general exercises as squat and knee extension) can positively condition the patellar stabilizers. Traditionally, some clinicians used a “specific” approach for PFPS, such as EMG biofeedback training and exercises targeting the unconditioned VMO muscle (Davlin et al 1999). According to the results of the study, the “global” approach of physical exercise seems to be as effective as VMO biofeedback training for improving the VMO/VL amplitude ratio. Since this study involved healthy subjects only, the effect of the training programs may not be generalized to subjects who belong to very different age groups or anthropometric types. For people with PFPS, the resistive exercises might need to be modified for rehabilitation purposes, such as the parallel squat that could lead to substantial patellofemoral and tibiofemoral compressive force that should be avoided for persons with anterior knee pain (Steinkamp et al 1993). Also, for people who need to avoid compressive force on their spines, the leg press machine is an alternative method to the barbell squat (Rogers 2001).

Jager and Luttman (1997) reported that the tolerance of cartilage against mechanical loading declines with aging. Thus, an aging patellofemoral joint would become less resistant to PFPS. Although no information on the long-term effect of physical conditioning for minimizing the PFPS incidence is available, it is logical to advocate the PFPS prevention exercise or “pre-habilitation” approach which should start before the patellofemoral joint degeneration is initiated. This is particularly important, for sportsmen who aspire to perform at the top level and maximize their athletic career life, the prevention program should start early.

The advantages of the present weight trainings are relatively low cost and technically simple, and the equipment used in the program are all available in most health clubs or gymnasiums worldwide. This would make it very easy for the study to be replicated in other population for other outcome measurements.

7.2. Adaptation of the passive and active patellar stabilizers

Increased passive lateral patellar mobility was repeatedly documented to be a contributing factor of the PFPS, patellar subluxation and dislocation in retrospective studies (Hughston et al 1984; Fithian et al 1995; Nomura et al 2006), prospective research (Witvrouw et al 2000) and review articles (Wilk et al 1998; Arendt et al 2002; Post 2005; Fredericson & Yoon 2006; Dixit et al 2007). The finding that lateral patellar mobility decreased after the weight training program in this study is in line with the results in Reeves et al (2003) that after 14 weeks of resistance training for previously untrained persons, their patellar tendon elongation decreased at a given isometric knee extension force as detected by a B-mode ultrasound. This suggests that the connective tissues bridging the patella to femur and tibia are able to adapt to mechanical loading. The major part of the adaptation might take place in the medial patellofemoral ligament (MPFL) that provides approximately 50% of resistance when the patella is passively translated to the lateral side (Conlan et al 1993; Hautamaa et al 1998).

The biomaterial properties of ligament are determined by collagen fiber contents, fiber orientation, and cross-link density (Ng et al 1995, 1996; Ng 2003; Fung et al 2003, 2004). These could be altered by weight training because mechanical loading is known to stimulate biosynthetic activity of the fibroblasts (Harris et al, 1981; Wren et al 1998; See et al 2004). If the conditioned ligament becomes stiffer after resistance training, the ligament should elongate less at a given force (Reeves et al 2003). Since the VMO muscle attaches to the medial one-third of the MPFL (Tuxoe et al 2002; Nomura et al 2005), when the VMO contracts, the MPFL is pulled proximally and tightened; therefore, the VMO not only acts as a directly active stabilizer of the patella but also indirectly through the MPFL (Nomura et al 2005). Also, the hardened MPFL could be more effective in force transmission between VMO and patella (Bojsen-Moller et al 2005), and could facilitate quicker and greater force development of the VMO on the patella in both static and dynamic actions.

By within-subject comparison of the surface EMG amplitude of VMO relative to VL, the 8-week weight training had increased the relative VMO electrical activity at a given functional task of sit-to-stand. The increased VMO/VL activity ratio signifies greater VMO recruitment and medial patellar force to counteract the lateral force produced by VL. The reason of this change could be theorized that weight training for previously untrained persons may serve as a learning process for adjusting the VMO and VL muscle recruitment so as to provide an effective position of the patella throughout its range of motion under functional or postural loading. The increasable VMO activation was also reported by Ng et al (2006) for subjects with PFPS after a 6-week biofeedback training.

Since the present study only captured the static MR images of the patellar position in supine lying with the whole body relaxed, the finding of magnified VMO/VL activity ratio may not directly imply that the patellar tracking or patellofemoral joint contact area changed after the interventions during dynamic actions. For treatment or prevention of PFPS, it may be essential to regulate the contact stress of a symptomatic patellofemoral joint (Powers et al 2004a, 2004b) and thus potential effect of the resistive training on the patellar tracking needs to be investigated further.

The VMO onset prior to that of VL is a pre-programmed motor control for the VMO to generate adequate muscle force to stabilize the patella; otherwise the VL may laterally displace the patella excessively (Neptune et al 2000). If VL fires before VMO frequently, the lateral patellofemoral joint may be overloaded resulting in anterior knee pain (Voight & Weider 1991; Grabiner et al 1994). In the present study, even though all the subjects were asymptomatic, not all of them were found to have their VMO onsets occurred prior to that of VL. This implies that the VMO-VL onset time difference is not a very specific indicator for the presence of PFPS. Also, the pre-post training improvement was approximately 4 ms in the intervention groups; it is not known whether the 4-ms difference can lead to significant change of the patellar tracking and patellofemoral joint contact area, although it was found in Neptune et al (2000) that stated a 5-ms difference in EMG onset between VMO and VL could affect the patellar tracking.

7.3. Knee joint proprioception

The knee joint position sense test was designed to measure the capability of a subject to precisely reproduce the preset target knee angle (Barrack et al 1983, Baker et al 2002). The active knee angle reproduction reflected the sensitivity of quadriceps and hamstrings muscle and knee joint capsular receptors, and the central nervous system's ability of close-loop efferent response after receiving the neural input from the muscles and the joint (Laskowski et al 2000). During the test, muscle spindles play a dominant role in the conscious perception of knee joint position as they detect changes in length of the intrafusal muscle fibers of quadriceps and hamstrings, also co-activation of gamma motorneurons which control the sensitivity of the spindle receptors, and the gamma motorneurons that are governed by the reticular system, cerebellum and basal ganglia of the central nervous system. All these structures chiefly contribute to the joint kinesthetic accuracy (McCloskey 1978; Kibler & Livingston 2001), and the 8-week weight training appears to have a positive effect on these neurological structures. Yet it is not known if this kinesthetic improvement would benefit the patellofemoral or knee joint stability during closed kinetic-chain and vertically loaded dynamic movements, such as running or landing action (Lephart et al 1998a; Williams et al 2001).

7.4. Limitations

The study introduced in Chapter 6 investigated the effect of two modes of weight training that lasted for 8 weeks. As a relatively short term training for previously untrained adults, the findings on strengthening patellar stabilizers may not be directly applicable for athletes who have had weight training experience, younger athletes or those with a history of PFPS.

Although females were more susceptible to the PFPS (Taunton et al 2002; Murray et al. 2005), the majority of the participants (75%) in the study introduced in Chapter 6 were males. This was due to the difficulties in recruiting women in Hong Kong for weight training as they tend to believe that weight training can easily and permanently make them look muscular. In fact, four out of five subjects withdrew during the study were females.

The present findings reveal that both modes of weight training were effective for VMO muscle building, but the potential change in the distribution of fast and slow twitch muscle fibers and the relative areas of these fibers (MacDougall et al 1980; Hostler et al 2001) have not been investigated. Changes in the muscle fiber composition or ratio could have an impact on the muscle shortening velocity or motor unit firing synchronization in VMO and VL that may affect the patellofemoral joint tracking or articular contact area during dynamic motions (Mellor & Hodge 2005).

7.5. Future study

Future study on potential effects of the resistance training for preventing PFPS should examine the change of patellar cartilage volume, dynamic patellar tracking (Powers et al 2004a, 2004b), vasti muscle fiber composition and muscle fiber orientation; those are quantitative factors related to PFPS or patellar biomechanical stability. In addition, isokinetic training, exercise on unstable surface, and different dosages, sequence, frequency of weight training, are all potential themes for future studies. Most importantly, a duration of the positive effects on the active and passive patellar stabilizers that can remain after training stops, and the possible long-term effect of resistance training on reducing risk of PFPS should be further examined.

7.6. Implications

Panjabi (1992) suggested that the stability of a movable articular joint was maintained by three systems, namely passive system constituted by connective tissues, active system formed by skeletal muscles, and neural system as a commander of the active system through motor control. A dysfunction of any one of the subsystems may lead to either the compensational adaptation of one or more subsystems, or overall dysfunction of articular joint stability. This theory may be applied to the patellofemoral joint; the weight training methods investigated in this thesis did positively affect these three systems including the active and passive patellar stabilizers and neuromuscular control of vasti muscles. There is evidence showing the change of EMG in the vasti muscles after training despite the magnitude of change was not large.

An important practical implication of this thesis is that both 8-week weight training targeted for muscle hypotrophy or strength were beneficial in attenuating the predisposing factors of PFPS and strengthening the patellar stabilizers for previously untrained individuals. The findings do reinforce the importance of physical conditioning, and can be regarded as additional benefits of weight training that were formerly unknown to clinicians or trainers. People who lack physical activities can not earn the effect of trainable patellofemroal stability. According to the results of this thesis, the conditioned patellar stabilizers were illustrated by enlarged VMO muscle, reduced passive lateral patellar mobility, increased knee extension force, greater recruitment and faster onset of VMO muscle in functional movement and enhanced knee joint position sense.

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Appendix 1

THE HONG KONG
POLYTECHNIC UNIVERSITY
香港理工大學

MEMO

To : NG Yin Fat, Department of Rehabilitation Sciences
From : KWONG Shek Chuen, Chairman, Departmental Research Committee, Department of Rehabilitation Sciences

Ethical Review of Research Project Involving Human Subjects

I write to inform you that approval has been given to your application for human subjects ethics review of the following research project for a period from 31/05/2005 to 01/12/2006:

Project Title : Different electrode placements on vastus lateralis lead to different characteristics of EMG?

Department : Department of Rehabilitation Sciences

Principal Investigator : NG Yin Fat

Please note that you will be held responsible for the ethical approval granted for the project and the ethical conduct of the research personnel involved in the project. In the case the Co-PI has also obtained ethical approval for the project, the Co-PI will also assume the responsibility in respect of the ethical approval (in relation to the areas of expertise of respective Co-PI in accordance with the stipulations given by the approving authority).

You are responsible for informing the Departmental Research Committee Department of Rehabilitation Sciences in advance of any changes in the research proposal or procedures which may affect the validity of this ethical approval.

You will receive separate notification should you be required to obtain fresh approval.

KWONG Shek Chuen
Chairman
Departmental Research Committee
Department of Rehabilitation Sciences

Appendix 2

The Hong Kong Polytechnic University
Department of Rehabilitation Sciences

Research Project Informed Consent Form

Project title:
Reliability of double peak-to-peak method for determining onset of vasti muscles as a potential physical diagnosis for anterior knee pain

Investigator(s):
Ng, Gabriel, PhD, Associate head, Dept. of Rehabilitation Sciences
Wong, Yiu Ming, MSc, PhD student, Dept. of Rehabilitation Sciences

Project information:
This pilot study will invite you to take part twice with one-week apart. The trials will take less than two hours and one hour respectively. You will need to wear sport shoes and a sport short to facilitate the measurements. After the skin of your thigh above knee joint being cleaned with alcohol and cleaning gel (a small area of your thigh may be shaved if necessary), a skin maker will draw few spots on the skin and three surface transducers will be placed on your thigh muscle for bioelectrical recording while you performing static kicking action in sitting position. Two minutes rest between those actions will be allowed. The procedures of this study will not result in any side effect and harm. However, you should report any discomfort encountered during the study to your investigator. The collected data in the trials will be used for research purpose only and your personal information will be strictly confidential.

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, Prof. Gabriel Ng at telephone 2766-6721 for any questions about this study. If I have complaints related to the investigator(s), I can contact Mrs. Michelle Leung, secretary of Departmental Research Committee, at 27665397. I know I will be given a signed copy of this consent form.

Signature (subject):__________________________Date:_____________________

Signature (witness): _________________________Date:_____________________

Appendix 3

THE HONG KONG
POLYTECHNIC UNIVERSITY
香港理工大學

MEMO

To : NG Yin Fat, Department of Rehabilitation Sciences
From : KWONG Shek Chuen, Chairman, Departmental Research Committee, Department of Rehabilitation Sciences

Ethical Review of Research Project Involving Human Subjects

I write to inform you that approval has been given to your application for human subjects ethics review of the following research project for a period from 31/05/2005 to 01/12/2006:

Project Title : Different electrode placements on vastus lateralis lead to different characteristics of EMG?

Department : Department of Rehabilitation Sciences

Principal Investigator : NG Yin Fat

Please note that you will be held responsible for the ethical approval granted for the project and the ethical conduct of the research personnel involved in the project. In the case the Co-PI has also obtained ethical approval for the project, the Co-PI will also assume the responsibility in respect of the ethical approval (in relation to the areas of expertise of respective Co-PI in accordance with the stipulations given by the approving authority).

You are responsible for informing the Departmental Research Committee Department of Rehabilitation Sciences in advance of any changes in the research proposal or procedures which may affect the validity of this ethical approval.

You will receive separate notification should you be required to obtain fresh approval.

KWONG Shek Chuen
Chairman
Departmental Research Committee
Department of Rehabilitation Sciences

Appendix 4

The Hong Kong Polytechnic University
Department of Rehabilitation Sciences

Research Project Informed Consent Form

Project title:
Different electrode placements on vastus lateralis lead to different characteristics of EMG?

Investigators: Gabriel Ng, PhD, Professor, Department of Rehabilitation Sciences
Wong Yiu Ming, MSc, BSc PT, CSCS, PhD student

Procedures of the study:
You are invited to attend one testing session for about one hour. You will need to wear sport shoes and shorts to facilitate the measurements. The skin of your thigh above the knee will be cleaned with alcohol and some gel (a small area of your thigh may be shaved if necessary), a few spots on the skin will be marked with skin markers and four surface electrodes will be placed on your thigh muscle for bioelectrical recording while you perform static kicking actions in sitting position. You will also be tested in standing on one leg and you will be perturbed at the back of your knee with a swinging pendulum weight.

Potential risk:
The procedures of this study would not result in any side effect or harm. However, you should report any discomfort encountered during the study to your investigator.

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 participating in 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 will not appear on any publications resulted from this study.

I can contact the chief investigator, Prof. Gabriel Ng at telephone 2766-6721 for any questions about this study. If I have complaints related to the investigator(s), I can contact Mrs. Michelle Leung, secretary of Departmental Research Committee, at 27665397. I know I will be given a signed copy of this consent form.

Signature (subject):_________________________Date:____________________

Signature (witness): ________________________Date:____________________

Appendix 5

THE HONG KONG
POLYTECHNIC UNIVERSITY
香港理工大學

MEMO

To : NG Yin Fat, Department of Rehabilitation Sciences
From : LEE Yun Wah Raymond, Chairman, Faculty Research Committee, Faculty of Health & Social Sciences

Ethical Review of Research Project Involving Human Subjects

I write to inform you that approval has been given to your application for human subjects ethics review of the following research project for a period from 04/ 04/ 2005 to 01/ 01/ 2008:

Project Title : Effects of weight training on patellofemoral joint position and quadriceps strength

Department : Department of Rehabilitation Sciences

Principal Investigator : NG Yin Fat

Please note that you are responsible for informing the Faculty Research Committee Faculty of Health & Social Sciences in advance of any changes in the research proposal or procedures which may affect the validity of this ethical approval.

You will receive separate notification should you be required to obtain fresh approval.

LEE Yun Wah Raymond
Chairman
Faculty Research Committee
Faculty of Health & Social Sciences

Appendix 6.1

The Hong Kong Polytechnic University
Department of Rehabilitation Sciences

Research Project Informed Consent Form

<u>Project title:</u>

Effect of weight training on patellar active and passive stabilizers

<u>Investigators:</u>

Gabriel Ng, PhD, Professor and Associate head, Dept of Rehabilitation Sciences, HKPU

Yiu-Ming Wong, MSc, PhD student, Dept of Rehabilitation Sciences, HKPU

<u>Procedures of the study</u>:

In agreeing to participate in this study, you will receive an 8 to 9-week guided weight training program at the HK Polytechnic University by a qualified fitness trainer for your thigh muscles. At the same time, you will be assessed three times at the HK Polytechnic University. Each assessment will last for about 90 minutes and you should wear sport shoes and shorts during the assessments which will include-

1) Body weight, height, body composition, thigh girth and lower body flexibility.
2) Magnetic resonance and/or ultrasound imaging for your lower thighs and knees while you lying supine and relaxed (the magnetic resonance imaging will last for 30 minutes each time and take place in the Queen Elizabeth Hospital).
3) Your knee cap will be gently moved and knee joint alignment will be measured while you lie on a plinth.
4) An angle measuring device will be put on one side your knee joint. With your eyes closed, your knee will be passively moved to different angles and then you will be asked to reproduce the angles on your own.
5) The skin of your thigh above the knee joint will be cleaned with alcohol and cleaning gel (a small area of your thigh may be shaved if necessary); four surface electrodes will be placed on your thigh muscle for recording the muscle activities when you perform some knee and lower limb movements.

All the procedures of this study will not result in any side effect. However, you should report any discomfort encountered during the study to your investigator. The collected data in the study will be used for research purpose only and your personal information will be strictly confidential.

Benefits to the subjects and to the society:

The benefits of participating in this study are learning the resistive exercises that should help you to enhance your physical fitness under professional supervision, and making contribution to the development of a knee pain prevention protocol for general public.

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, Prof. Gabriel Ng at telephone 2766-6721 for any questions about this study. If I have complaints related to the investigator(s), I can contact Mrs. Michelle Leung, secretary of Departmental Research Committee, at 27665397. I know I will be given a signed copy of this consent form.

Signature (subject): ______________________________ Date: ______________________

Signature (witness): ______________________________ Date: ______________________

Signature (investigator): ___________________________ Date: _____________________

Appendix 6.2

香港理工大學

康復治療科學系

科研同意書

科研題目： 阻力訓練對髕股關節穩定肌之影響

科研人員： 吳賢發教授 (康復治療科學系副主任, 香港理工大學)

黃耀明先生 (康復治療科學系博士研究生, 香港理工大學)

科研內容：如閣下同意參與此研究，將會在香港理工大學內及合格教練指導下接受 8 至 9 星期之下肢阻力訓練，期間會進行 3 次測試，每次約 1½小時，請準備運動鞋及短褲，測試包括如下：

1) 體重、身高、脂肪比例，大腿圍及下身柔軟度。

2) 在仰臥情況下，磁力共振 及/或 超音波影像測量大腿及膝關節 (磁力共振測量每次時間為半小時，地點在伊利沙伯醫院)。

3) 在仰臥情況下，量度膝蓋骨及膝關節之移動度及排列角度。

4) 一量角器會貼於閣下膝旁，在你雙眼閉上時，閣下之膝關節會被動地移至不同角度，稍後你需要主動地重覆該膝關節角度。

5) 閣下之大腿肌肉近膝部會被用酒精清潔(必要時少量毛髮需要剃去)，4 個傳感器會貼於大腿上量度不同動作時之肌肉電流。

所有研究之項目都沒有副作用，但如果你在研究期間感到任何不適，請告知科研人員。而收集之數據只會用作研究用途，閣下之個人資料會絕對保密。

<u>對項目參與人仕和社會的益處</u>：參與此研究之效益爲，閣下可以在專人指導下學習阻力訓練，以幫助改善體適能，並爲發展一套防止膝痛之計劃作出貢獻。

<u>同意書</u>：

本人____________已瞭解此次研究的具體情況。本人願意參加此次研究, 本人有權在任何時候、無任何原因放棄參與此次研究，而此舉不會導致我受到任何懲罰或不公平對待。本人明白參加此研究課題的潛在危險性以及本人的資料將不會洩露給與此研究無關的人員，我的名字或相片不會出現在任何出版物上。

本人可以用電話 2766-6721 來聯繫此次研究課題負責人 吳賢發教授。若本人對此研究人員有任何投訴，可以聯繫梁女士（部門科研委員會秘書），電話：27665397。本人亦明白，參與此研究課題需要本人簽署一份同意書。

簽名（參與者）: ______________________ 日期：____________________

簽名（證人）: ______________________ 日期：____________________

簽名（科研人員）: ____________________ 日期：____________________

Appendix 6.3

The Hong Kong Polytechnic University
Department of Rehabilitation Sciences

Research Project Informed Consent Form

Project title:
Effect of weight training on patellar active and passive stabilizers

Investigators:
Gabriel Ng, PhD, Professor and Associate head, Dept of Rehabilitation Sciences, HKPU
Yiu-Ming Wong, MSc, PhD student, Dept of Rehabilitation Sciences, HKPU

Procedures of the study:
In agreeing to participate in this study, you will be requested not to alter your regular diet and physical activities for 8 weeks. Also, you will be assessed twice at the HK Polytechnic University by a registered physical therapist at 8-week apart. Each assessment will last for about 90 minutes and you should wear sport shoes and shorts during the assessments which will include-

1) Body weight, height, body fat ratio, thigh girth and lower body flexibility.
2) Ultrasound imaging measurement for your lower thigh size while you lie on a plinth.
3) Your knee cap will be gently moved and knee joint alignment will be measured while you lie on a plinth.
4) An angle measuring device will be put on one side your knee joint. With your eyes closed, your knee will be passively moved to different angles and then you will be asked to reproduce the angles on your own.
5) The skin of your thigh above the knee joint will be cleaned with alcohol and cleaning gel (a small area of your thigh may be shaved if necessary); four surface electrodes will be placed on your thigh muscle for recording the muscle activities when you perform some knee and lower limb movements.

All the procedures of this study will not result in any side effect. However, you should report any discomfort encountered during the study to your investigator. The collected data in the study will be used for research purpose only and your personal information will be strictly confidential.

Benefits to the subjects and to the society:

The benefits of participating in this study are making contribution to the development of a knee pain prevention protocol for general public.

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, Prof. Gabriel Ng at telephone 2766-6721 for any questions about this study. If I have complaints related to the investigator(s), I can contact Mrs. Michelle Leung, secretary of Departmental Research Committee, at 27665397. I know I will be given a signed copy of this consent form.

Signature (subject): ___________________________ Date: ______________________

Signature (witness): ___________________________ Date: ______________________

Signature (investigator): ________________________ Date: _____________________

Appendix 6.4

香港理工大學

康復治療科學系

科研同意書

科研題目： 阻力訓練對髕股關節穩定肌之影響

科研人員： 吳賢發教授 (康復治療科學系副主任, 香港理工大學)

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科研內容：如閣下同意參與此研究，將會在香港理工大學內接受進行相隔 8 星期之 2 次測試，每次約 1.5 小時，請準備運動鞋及短褲，測試包括如下：(在 8 星期期間，請避免改變運動及飲食習慣)

1) 體重、身高、脂肪比例，大腿圍及下身柔軟度。
2) 在仰臥情況下，超音波影像量度大腿肌肉。
3) 在仰臥情況下，量度膝蓋骨及膝關節之移動度及排列角度。
4) 一量角器會貼於閣下膝旁，在你雙眼閉上時，閣下之膝關節會被動地移至不同角度，稍後你需要主動地重覆該膝關節角度。
5) 閣下之大腿肌肉近膝部會被用酒精清潔(必要時少量毛髮需要剃去)，4 個傳感器會貼於大腿上量度不同動作時之肌肉電流。

所有研究之項目都沒有副作用，但如果你在研究期間感到任何不適，請告知科研人員。而收集之數據只會用作研究用途，閣下之個人資料會絕對保密。

<u>**對項目參與人仕和社會的益處**</u>：參與此研究之效益爲，爲發展一套防止膝痛之計劃作出貢獻。

<u>同意書</u>：

本人＿＿＿＿＿＿已瞭解此次研究的具體情況。本人願意參加此次研究, 本人有權在任何時候、無任何原因放棄參與此次研究，而此舉不會導致我受到任何懲罰或不公平對待。本人明白參加此研究課題的潛在危險性以及本人的資料將不會洩露給與此研究無關的人員，我的名字或相片不會出現在任何出版物上。

本人可以用電話 2766-6721 來聯繫此次研究課題負責人 吳賢發教授。若本人對此研究人員有任何投訴，可以聯繫梁女士（部門科研委員會秘書），電話：27665397。本人亦明白，參與此研究課題需要本人簽署一份同意書。

簽名（參與者）: ＿＿＿＿＿＿＿＿＿＿＿＿ 日期：＿＿＿＿＿＿＿＿＿＿＿

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簽名（科研人員）: ＿＿＿＿＿＿＿＿＿＿＿ 日期：＿＿＿＿＿＿＿＿＿＿＿

Appendix 7

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Physical Therapy in Sport 7 (2006) 122–127

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Original research

Surface electrode placement affects the EMG recordings of the quadriceps muscles

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Received 3 November 2005; received in revised form 17 March 2006; accepted 27 March 2006

Abstract

Objectives: This study examined the EMG activities of the medial and lateral vasti muscles with four different surface electrode positions during isometric knee extension and knee perturbation.

Design: Repeated measures design.

Setting: Orthopaedic rehabilitation laboratory of a university.

Participants: Eight able-bodied non-athletic male volunteers.

Main outcome measures: The relative onset time difference in milliseconds of vastus medialis obliquus (VMO) and vastus lateralis (VL) during knee perturbation, and the normalized EMG amplitude in root-mean-square ratio of VMO:VL during submaximal isometric knee extension were analyzed using repeated measures ANOVA.

Results: Different electrode positions on VL resulted in different VMO-VL onset time ($p = 0.002$) and normalized VMO:VL activity ratios ($p = 0.002, 0.014$).

Conclusions: The position of surface electrodes has significantly affected the EMG readings of the vasti muscles. This finding has vital clinical implications for the application of EMG measurement and biofeedback training for the rehabilitation of the vasti muscles.

Keywords: Electromyography; Electrode; Knee; Quadriceps muscle

1. Introduction

Patellofemoral pain syndrome (PFPS) presents as localized pain, inflammation, muscle imbalance, and instability of any component of the knee extensor mechanism (LaBotz, 2004). The PFPS affects approximately 25% of the population at some stage in their lives (McConnell, 1996) with females being more vulnerable than males (Almeida, Trone, Leone, Shaffer, Patheal, & Long, 1999).

Patients with PFPS usually respond to non-operative treatments and the outcome is usually assessed with the level of pain, patellar alignment, soft-tissue flexibility, muscle activation, and coordination (Post, 2005). The vastus medialis obliquus (VMO) and vastus lateralis (VL) muscles have different morphological, functional, and neuromotor characteristics (Crossley, Bennell, Green, & McConnell, 2001; Lieb & Perry, 1968; Ng, 2002). The two muscles work synergistically to provide the dynamic stabilizing force on the patella during knee extension and imbalance between the VMO and VL would disturb the dynamic stability of patella. Clinically, weakness of the VMO would result in excessive lateral tracking of patella, which has been regarded as a common cause of PFPS (Herrington, 1998; McConnell, 1996). Furthermore, some studies had reported that persons with PFPS had reduced VMO:VL electrical

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doi:10.1016/j.ptsp.2006.03.006

activity ratio and delayed VMO onset when compared with normal controls (Mariano & Caruso, 1979; Souza & Gross, 1991; Tang, Chen, Msu, Chou, Mong, & Lew 2001; Voight & Wieder, 1991; Witvrouw, Sneyers, Lysens, Victor, & Bellemans, 1996).

The relative activity of VMO and VL has been studied in the past. Souza and Gross (1991) reported the normalized VMO:VL activity ratio in healthy subjects was approximately 1:1 during isometric knee extension, whereas Boucher, King, Lefebvre, and Pepin (1992) reported the ratio to be approximately 2:1 for asymptomatic subjects in static terminal knee extension. Tang et al. (2001) demonstrated that the ratio was less than 1 for normal subjects at the end range of isokinetic knee extension.

In terms of the temporal parameters of the vasti muscles, Voight and Wieder (1991) and Witvrouw et al. (1996) reported that VL fired significantly earlier than VMO in the knee jerk reflex in patients with PFPS, which was in contrast to the findings of Karst and Willett (1995) who reported no difference in VMO-VL contraction timing in the knee jerk reflex in both normal subjects and those with PFPS. Powers, Landel, and Perry (1996) also found that the VMO:VL activity ratio and the VMO-VL relative onset time were not different between patients with PFPS and the normal controls during walking or climbing stairs.

Some researchers had attempted to develop specific strengthening exercises for the VMO. Hanten and Schulthies (1990) and Laprade, Culham and Brouwer (1998) reported that VMO was more active than VL during isometric knee extension with hip adduction and tibial internal rotation. Lam and Ng (2001) and Ng and Man (1996) reported that VMO was significantly more active than VL with some knee flexion and internal hip rotation. However, Mirzabeigi, Jordan, Gronley, Rockowitz, and Perry (1999) and Reynolds, Levin, Medeirosm, Adler, and Hallum (1983) did not support the concept of selective recruitment of VMO over VL in physical training.

The discrepancies of the above findings could be due to the electrode placements not being comparable among these studies. Boucher et al. (1992) placed an electrode over the motor point of VL at the mid-thigh level of 15 cm above the upper border of patella. Gilleard, McConnell, and Parsons (1998) placed the VL electrode at 10 cm above the patella. Cram and Kasman (1998) recommended that the electrode for the VL muscle should be at 5 cm above the knee. Conversely, the electrode placement for VMO was more consistent among previous studies as most reported the position to be 3 or 4 cm above the supero-medial corner of patella (Cram & Kasman, 1998; Gilleard et al., 1998). In light of the diverse positioning of the VL electrode and the lack of information on the effects of surface electrode positioning on the EMG recording, there is a need to examine if the electrode positioning would affect the EMG measurements. This has vital clinical implications because imbalanced VMO:VL EMG activity ratio or VMO-VL onset time are regarded as contributing factors of PFPS, the effect of electrode positioning must be considered in order to make accurate assessments. Therefore, the present study was conducted to investigate the effects of different surface electrode placements on the VMO:VL magnitude ratio and VMO-VL relative onset timing.

2. Methods

2.1. Subjects

Eight able-bodied non-athletic males aged between 24 and 35 years (mean 28.1 years) were recruited. Subjects with history of lower limb operations or injuries that required treatments in the past 6 months were excluded. The study was reviewed and approved by the Human Ethics Sub-committee of The Hong Kong Polytechnic prior to data collection. All subjects were asked to refrain from vigorous physical activities the day before the study.

2.2. EMG recordings

Skin on the distal thigh of the dominant leg was shaved and cleansed with methylated spirit and skin preparation gel (NuPrep, DO Weaver & Co., Aurora, CO, USA). The skin impedance was checked with an impedance meter (1089MKIII, UFI, Morro Bay, CA, USA) and a value of less than 50 kΩ was deemed acceptable (Hewson, Hogrel, Langeron, & Duchene, 2003).

After the skin preparation four single-differential EMG electrodes (Myoscan, Thought Technology, Montreal, Que., Canada) were placed over the VMO and VL with disposable/adhesive circular snap electrodes (Multi Bio Sensors, El Paso, TX, USA). The electrode for VMO was placed at a point 4 cm above the supero-medial corner of patella, whereas three electrodes were placed over VL at 5, 10, and 15 cm above the supero-lateral corner of patella (Fig. 1). The orientations of the VMO and VL electrodes were 55° and 15° to the longitudinal axis of femur, respectively (Lieb & Perry, 1968).

The electrodes were connected to a battery-powered encoder (SA9404P, Thought Technology, Montreal, Que., Canada) with a sampling rate of 1984 Hz. The encoder was fed to a digital signal processing board (DSP, version 2, Thought Technology, Montreal, Que., Canada) installed on a personal computer. Data acquisition software (V1.52 Ergonomic Suite, Thought Technology, Montreal, Que., Canada) was used to

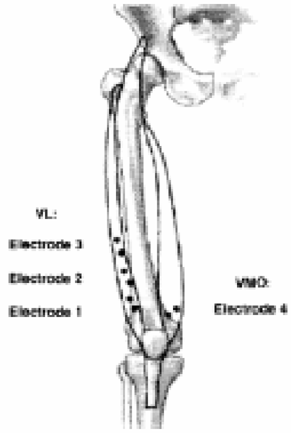

Fig. 1. Positioning of the EMG electrodes for VL and VMO of the right leg.

capture the raw EMG data with a bandwidth of 20–500 Hz. The raw data were later analyzed off-line with Excel software (Microsoft, Redmond, WA, USA).

2.3. *Isometric knee extension*

A dynamometer (Cybex Norm, Henley Healthcare, Nauppauge, NY, USA) was used to measure the isometric maximal voluntary contraction (MVC) on the dominant leg of each subject with the knee at 45° and hip at 85° of flexion. Subjects performed 5 repetitions of submaximal knee extension to get acquainted with the isometric testing prior to the MVC test that lasted for 4 s and was repeated 3 times with a 2-min rest in-between. The highest force developed in the 3 tests was regarded as the MVC. Afterwards, the subjects performed 5 s of isometric knee extension at 50% and 75% of their MVC with real-time visual signals from the dynamometer screen to guide them on the force output. These submaximal tests were implemented once since the EMG amplitude measurement with submaximal muscle contraction was confirmed highly repeatable in short-term (Kollmitzer, Ebenbichler, & Kopf, 1999). During the isometric testing, EMG activities of VMO and VL were recorded with the above data acquisition system and computer.

2.4. *Knee perturbation test*

After the knee extension tests, the subjects proceeded to a knee perturbation test with single leg standing so as to induce a stretch reflex in the quadriceps muscles (Ng, 2005). Subjects stood on their dominant leg with the knee extended and the hands lightly touching the wall in front for balance. A medicine ball of 3.5 kg was used to provide perturbation to the knee joint. The ball was suspended with a sling from the ceiling so that it could be adjusted to be at the level of the knee joint of the subjects. The examiner pulled the ball back until the sling made an angle of 45° to the vertical behind the subject and upon release of the ball, it would exhibit a pendulum swing and hit on the back of the subject's knee to produce a perturbation. The perturbation test was repeated 3 times with 1 min in-between while the EMG of VMO and VL were recorded.

2.5. *Data analysis*

For the isometric knee extension, the EMG signal was transformed to root-mean-square (RMS) in time constant of 1000 ms. The initial 2 s of EMG data were excluded so as to avoid the fluctuation of EMG signal. Then each subject's highest RMS values of VMO and VL at 50% and 75% MVC were normalized against the respective muscle's 100% MVC (Soderberg & Knutson, 2000). Since there were 3 VL electrodes and 1 VMO electrode, 3 normalized VMO:VL ratios were obtained for each MVC testing condition.

For the knee perturbation test, the EMG data were full-wave rectified and the mean and standard deviations of the 100-ms resting EMG signal prior to perturbation were calculated. The EMG onset was determined as a signal that had 2 standard deviations above the resting mean for 10 ms or more (McKinley & Pedotti, 1992). Since there were 3 electrodes on the VL, 3 VMO-VL relative EMG onset times were obtained. The mean EMG onset time of the 3 perturbation trials was calculated for each subject in order to minimize random error and intraclass correlation coefficient (ICC, absolute agreement, two-way mixed, CI 95%) was calculated to determine its within-day reliability.

Repeated measures ANOVA ($p = 0.05$) with Bonferroni adjustment for within-subject comparisons (SPSS 11.5.1, SPSS Inc., Chicago, IL, USA) was used to analyze the normalized VMO:VL activity ratio and the relative VMO-VL onset timing differences among the 3 different pairs of the VMO and VL electrodes.

3. Results

The ICC values of the EMG onset timing difference of VMO-VL in the knee perturbation measurement were 0.86, 0.87, and 0.87 for the electrode pairs of 1 and 4, 2 and 4, and 3 and 4, respectively. These ICCs indicated that the EMG recordings had good reliability (Portney & Watkins, 2000). The means and standard deviations of the VMO-VL onset time and VMO:VL electrical activities are listed in Table 1.

Table 1
The mean (standard deviations) of EMG onset timing during perturbation and EMG activity ratios of VMO:VL during 50% and 75% of MVC testing

Electrode pair	VMO-VL onset time (ms)	Normalized VMO:VL ratio	
		50% MVC	75% MVC
1 & 4	6.30 (3.35)	1.19 (0.04)	1.24 (0.07)
2 & 4	8.91 (4.96)	0.97 (0.08)	1.05 (0.08)
3 & 4	12.28 (7.10)	0.90 (0.09)	0.98 (0.08)

Table 2
The results of repeated measures ANOVA with Bonferroni adjustment of EMG onset timing during perturbation and EMG activity ratios of VMO:VL during 5% and 75% of MVC isometric knee extension

Measurements	ANOVA	Pair-wise comparison
VMO-VL onset in knee perturbation	df = 2, $F = 10.555$, $p = 0.002$	Electrode pair of 1:4<2:4 ($p = 0.03$) and 3:4 ($p = 0.004$)
VMO:VL ratio in 50% MVC	df = 1.028, $F = 21.157$, $p = 0.002$	Electrode pair of 1:4>2:4 ($p = 0.018$) and 3:4 ($p = 0.004$); 2:4>3:4 ($p = 0.000$)
VMO:VL ratio in 75% MVC	df = 1.044, $F = 10.121$, $p = 0.014$	Electrode pair of 1:4 and 2:4>3:4 ($p = 0.018$, 0.004)

Results of repeated measures ANOVA revealed significant difference in both the VMO:VL ratios and VMO-VL onset time (Table 2). Post hoc analyses revealed that the recording with electrode pair 1:4 was significantly different from the electrode pairs of 2:4 and 3:4 for the onset timing. For the VMO:VL activity ratio, electrode pair 1:4 had higher ratios than 2:4 and 3:4 at 50% activation level, whereas at 75% of activation level, both 1:4 and 2:4 had higher ratios than 3:4 (Table 2).

4. Discussion

This study aimed to examine the effect of electrode positioning on the EMG recordings of VMO and VL with respect to the onset timing during quadriceps stretch reflex and voltage during voluntary isometric contraction. Results revealed that different positioning of the VL electrodes would significantly affect the EMG onset time and electrical signal strength during the same activity. The present findings may explain the discrepancies in the previous studies that reported inconsistent VMO:VL activity ratio and VMO-VL onset timing (Boucher et al., 1992; Karst & Willett, 1995; Laprade et al., 1998; Mirzabeigi, et al., 1999; Souza & Gross, 1991; Tang et al., 2001; Voight & Wieder, 1991; Witvrouw et al., 1996). Our results suggested that the EMG signals obtained from the electrode closer to the VL innervation zone at mid-thigh level (Warfel, 1985) had lower electrical strength and delayed onset time. These findings are comparable to the reports of Boucher et al. (1992) and Souza and Gross (1991). The former study placed the electrode on mid-VL and reported the VMO:VL ratio to be 2:1, while the latter study positioned the electrode on the distal VL and reported the ratio to be 1:1, both in terminal isometric knee extension.

The EMG electrodes used in this study contained a differential amplifier with two recording and one reference terminals. The differential amplifier would magnify the non-identical signals between the two input terminals and reject any signal which is common to both (Peek, 1987). Theoretically, no EMG signal can be detected if the two recording terminals are placed equally on both sides of the motor end-plate of a muscle along its fiber direction, because when a muscle action potential is generated at the motor end-plate and propagates to both ends of the muscle, the signals will be registered as identical by both input thus cancelled by the differential amplifier (Campbell, 1999). However, this will not happen with in vivo measurement, because the electrodes detect muscle action potentials generated by different motor end-plates instead of from a single end-plate. These signals will reach the input terminals at slightly different time intervals thus complete cancellation will not occur. However, due to the phasic proximity of the signals, partial cancellation can happen which may result in smaller EMG amplitudes. Conversely, electrodes located midway between the motor points and tendinous insertion would result in relatively larger signal amplitude since the action potentials would reach the input terminals with very different phasic properties (Kleine, Stegeman, Mund, & Anders, 2001). The EMG amplitudes of upper trapezius and tibialis anterior muscles have been reported to be relatively lower when the electrodes were placed near the innervation zones during both voluntary contraction and electrical stimulation (Jensen, Vasseljen,

& Westgaard, 1993; Merletti, LoConte, Cisari, & Massazza, 1993). This may explain our present findings that electrode 3 located in the middle of VL had the lowest signal strength because it is where most of the motor end-plates are found and they cluster around this electrode position (Warfel, 1985).

For the EMG onset timing, results revealed that position 3 had a significantly later onset time than the other two positions. This was in agreement with the finding on the voltage because the determination of the onset was dependent on the difference between the resting discharge voltage and that of the action potential signal strength. With a relatively lower action potential strength detected in position 3, the rate of voltage build-up in this position would also be slower, thus the onset point was registered later.

Our findings indicated that the activity ratio and onset timing for the VMO and VL were electrode position dependent. Studies using unique electrode locations over the vasti muscles could result in incompatible outcomes. However, the finding is restricted to surface electrodes, it is not known if the same finding can be applied to wire EMG as the wire EMG directly detects the individual motor unit action potential rather than multiple motor unit action potentials monitored on the skin in the surface EMG (Basmajian & DeLuca, 1985).

One of the limitations in the present study was a lack of external validity because only asymptomatic subjects were tested, thus the findings may not be directly applied to people with PFPS. In addition, all subjects' knee joints were either positioned at 45° flexion or fully extended, it is natural that if the knee joint flexes by more than 45°, the vasti muscles will be stretched further, thus the innervation zone of VMO and VL may shift inferiorly and the EMG characteristics may change accordingly.

In the present study, the variability of subjects' height was less than 8 cm. Thus the electrodes being positioned on a few spots on the quadriceps with standardized distances from the patella may not be a confounding factor for the EMG recording. However, for clinical measurements, the subjects' height could vary significantly; the use of a percentage of limb length may ensure the standardization of the electrode placements for individuals. This may also enhance between-subject and between-study comparisons.

5. Conclusion

The positions of surface electrodes would affect the EMG recording of VMO and VL. Physical therapists or researchers should be aware that a difference of a few centimeters in the position of the VL electrode could affect the VMO:VL EMG activity ratio and VMO-VL onset timing.

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Appendix 8

The double peak-to-peak analysis for determining EMG onset of muscle contraction

Y.M. Wong, G.Y.F. Ng

Abstract

Purpose: *Identification of the onset of muscle contraction with EMG signal amplitude double of the baseline value (DP-P) has been recently reported for determining the temporal parameters of muscular activity. Due to its convenience, it is suitable for clinical application. However, there is a lack of report on the reliability and comparability of this method to other established methods. Therefore, this study examined the test-retest reliability of the DP-P method and compared it with an established method that used the mean + 3 standard deviations (mean + 3 s.d.) over the baseline value for muscles of the knee.*

Methods: *The onset of contraction of vastus medialis obliquus (VMO) and vastus lateralis (VL) of eleven able-bodied volunteers performing isometric knee extension at 50%, 75% and 100% of MVC in 30-minute and 7-day intervals were analyzed with both the DP-P and mean + 3 s.d. methods.*

Results: *The ICC for within-day measurements of DP-P method ranged from 0.64 to 0.86 and that for between-day measurements ranged from 0.63-0.81. The ICC values were higher with submaximal than maximal contractions. There was a consistent delay of about 3 ms in EMG onset detection with the DP-P as compared to the mean + 3 s.d. method.*

Conclusion: *The DP-P method is a reliable method for muscle onset determination but the absolute onset time of muscle contraction obtained from this method should not be directly compared with other methods such as the mean + 3 s.d.*

Introduction

Identification of the onset of skeletal muscle activities is often required in the treatment of musculoskeletal disorders, muscle training and gait analysis. For determining the onset of muscle contraction, most clinicians rely on their visual inspection, palpation and experience. However, these observations are rather subjective and their reliability is often doubtful (1). With the need of improving the objectivity and reliability of muscle onset determination, researchers have used electromyography (EMG) to determine the onset of muscle activities. In order to accurately quantify the muscle onset time, various quantitative and computerised methods of analyzing the EMG signals have been developed in the past two decades. One of the most popular methods for determining the EMG onset was described by Hodges & Bui (2), in which the EMG signals were processed with 50Hz low-pass filter and rectification, the time of muscle onset was set as the instance that the EMG signal was at three standard deviations above the mean (mean + 3 s.d.) of the baseline value. This method has since been widely used by various researchers for studying different muscles under various testing modes (3-12).

However, analysis of the EMG onset time with this method requires the original EMG signals to be exported to a computer for mathematical processing, which may not be convenient for clinicians to perform in clinical settings. Furthermore, many EMG systems used by clinicians would only display the EMG waveforms thus any mathematical

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processing of the data could not be done. Therefore, a quantitative, reliable and convenient method for determining the muscle onset time is needed for easy application in the clinical settings.

Vaes et al. (13, 14) were the first group of researchers to report the EMG onset time of the peroneal muscle by using the amplitude of the surface EMG signals at double of that of the baseline values (DP-P). This method does not involve sophisticated EMG signal processing techniques, thus therapists can trace the EMG output signals with a caliper of the EMG monitor to locate the onset point. Therefore, it can be easily implemented in almost any EMG system that does not have statistics or data exporting functions.

However, the reliability of the DP-P method and the compatibility of the results of this method with other established methods such as the mean + 3 s.d. method have not been established. Therefore, this study aimed to (1) examine the reliability of identification of EMG onset using the double peak-to-peak method (DP-P), and (2) compare the results of this method with that of the mean + 3 s.d. method for the muscles of the knee in able-bodied subjects. If the DP-P method of muscle onset detection was found to be reliable and the results were comparable with the other method, it could be applied clinically for studying the muscle activation of subjects with different pathologies.

Materials and Methods

Subjects

Eleven subjects (five females and six males) with mean age of 29.4 years (+/- 8.44 years) volunteered to participate in this study. All subjects were healthy with no history of lower limb operations or injuries that required medical treatments. All subjects were asked to abstain from vigorous physical activities on the day before the study.

Instruments

Isometric dynamometer: An isokinetic dynamometer (Cybex Norm, Henley Healthcare, Nauppauge, NY, USA) was used to measure the isometric knee extension strength of the subjects' left leg at 45° of knee flexion and 85° of hip flexion. The knee extension strengths expressed in torque values were converted to DC voltages (1 V = 80 Nm) and transmitted to a battery powered encoder (SA9404P, Thought Technology, Montreal, Quebec, Canada) at sampling rate of 124 Hz via a voltage isolator (T9405, Thought Technology, Montreal, Quebec, Canada). The encoder was also connected to a digital signal processing board (DSP, version 2, Thought Technology, Montreal, Quebec, Canada) installed in a personal computer (Celeron 400MHz, Window 98). Data acquisition and analysis software (V1.52 Ergonomic Suite, Thought Technology, Montreal, Quebec, Canada) was used to provide real-time visual feedback of the torque output to subjects and for off-line data analyses.

EMG recording system: Two single-differential surface electrodes (SA9503M, Thought technology, Montreal, Quebec, Canada) were connected to the encoder for EMG data capturing. The bandwidth was set at 10-500 Hz and the sampling rate for the EMG data was 1984 Hz.

Procedures

This study was conducted in a laboratory and the procedures were reviewed and approved by the ethics review committee for human experimentation of our university. Only one subject was tested at a time. After all the procedures of the EMG measurement were explained and demonstrated to the subjects, the skin over the mid-point of vastus medialis obliquus (VMO) and vastus lateralis (VL) of the subject's left knee would be prepared by shaving and cleansing with skin preparation gel (NuPrep, DO Weaver & Co., Aurora, CO, USA) and alcohol.

The electrode placement was based on the report of Cram (15). The electrodes for VMO were positioned 4 cm superior to the superomedial patellar border and orientated 55° to the longitudinal femoral axis, whereas the electrodes for VL recording were positioned 8 cm superior and 2 cm lateral to the superolateral patellar border and orientated 15° to the femur. The electrodes were secured on VMO and VL by disposable circular snap electrodes (Multi Bio Sensors, El Paso, TX, USA) with a standard 2 cm spacing of silver/silver chloride electrodes that have pickup areas of 1cm in diameter. A thin layer of conductive gel (Ultra/Phone, Pharmaceutical Innovations, Newark, NJ, USA) was put on the pickup surfaces.

After the above preparations, each subject performed 3 maximal isometric knee extensions. The highest isometric extensor torque of the 3 contractions would be registered as the maximal voluntary contraction (MVC) of the subject. After registering the MVC, the subject would be given 2 minutes of rest and then the subject would perform the following actions in random order:

1) Explosive 100% MVC knee extension.
2) Explosive 75% MVC knee extension.
3) Explosive 50% MVC knee extension.

Visual feedback was provided to the subjects to help them to develop the required levels of MVC. A value of +/-10% around the targeted level of MCV was regarded as acceptable for these trials. All subjects were tested three times. The first two measurements were recorded with a 30-minute interval without removing the electrodes whereas the third measurement was recorded at 7 days later.

Data analysis

The sudden increase in voltage resulted from the isometric knee extension torque produced by the subject was regarded as the mechanical onset. An EMG signal of 200-ms in duration that preceded the mechanical output was manually selected as the baseline reference. Muscle activation onset point determined by the DP-P method was regarded as the point at which the EMG peak-to-peak intensity doubled that of the quietest peak-to-peak waveform in the baseline reference (13, 14, 16). After the identification of the EMG onset with the DP-P method, the raw EMG signal was low-pass filtered with 50 Hz and full wave-rectified. The onset point was determined as the instance at which the processed signal was higher than 3 standard deviations over the mean value of the reference signal and such signal lasted for 25 ms or more (2-8, 11).

Results

Intra-class Correlation Coefficients (ICC: absolute agreement, two-way mixed, CI 95%) was used to establish the test-retest reliability of the muscle onset time of the vasti muscles being determined by the DP-P (Table 1). A trend of increasing in test-retest reliability with decreasing in levels of MVC was noted.

Comparison of the DP-P to the mean + 3 s.d. was done using paired t-test with α set at 0.05. Significant differences were found between the results of the two methods for both VL and VMO ($p < 0.0001$). In the cell point chart that showed the means (standard deviation) of the EMG onset time of VMO and VL (Fig. 1), it can be seen that there was a consistent delay in the DP-P method as compared to that of the mean + 3 s.d. method.

Discussion

This study aimed to examine the test-retest reliability of the DP-P method for EMG onset identification and to compare the results of DP-P method with the mean + 3 s.d. method. Our findings revealed that DP-P method has moderate to good reliability with ICC ranging from 0.63-0.86 (17). Since EMG measurement is susceptible to surrounding electromagnetic noise that can substan-

Table 1. – *The ICC values indicating the reliability of the double peak-to-peak (DP-P) method*

	100% MVC		75% MVC		50% MVC	
Interval	VMO	VL	VMO	VL	VMO	VL
30-min	0.66	0.64	0.76	0.67	0.86	0.75
7-day	0.66	0.63	0.66	0.67	0.80	0.81

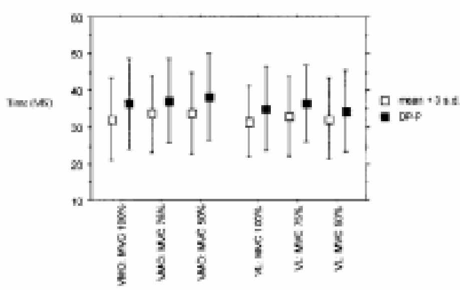

Fig. 1. – Comparison of onset time of contraction of VMO and VL determined by the mean + 3 s.d. and the DP-P methods.

tially increase the EMG offset, statistics such as the Pearson correlation is considered not a suitable index of reliability when there is substantial difference between two EMG measurements among the subjects and their ranks remain the same in the two measurements. ICC was used since it can detect both changes in the rank among the subjects tested and the consistency of the measurements across occasions (18).

The DP-P method showed that the EMG onset detection was more repeatable in the 50% MVC isometric knee extension than higher levels of MVC. This finding is in line with the report of Kollmitzer et al. (19) in which recordings of root-mean-square EMG amplitude of isometric knee extension taken at 50% MVC demonstrated better repeatability than 100% MVC measurements. The good reliability of this method with submaximal recording is clinically meaningful, because most of the therapeutic exercises or assessments of the vasti muscles are done at submaximal levels, and the EMG recordings at these submaximal contractions could be compared periodically.

More importantly, our results revealed very good between-day reliability of the DP-P method that the ICC for between-day comparisons was > 0.8 at 50% MVC. This finding implied that the effects of exercise training to facilitate muscle onset may be assessed with the DP-P method, which has important clinical application value.

VMO and VL are two muscles that their onset times have been widely explored with the mean + 3 s.d. method in previous studies (3-8, 11), thus the validity of the standard upon which the DP-P is compared to is well established. Furthermore, accurate measurements of the onset time of VMO and VL have vital clinical implications to the management of patients with patellofemoral pain syndrome as the treatment for these patients often targets to facilitate VMO contraction, developing a clinician friendly detection method will facilitate the clinical management of this condition.

However, a point of caution is that the findings on the vasti muscles may not be generalised to other skeletal muscles, due to surface EMG recordings could be affected by factors such as electrode placement, cross talk of adjacent muscles, muscular length and fatigue. In particular, muscles of the limbs work against inertia whereas other muscles of the body such as the chest wall work against elasticity and their EMG/force relationships were found to be different (20). Applications of the findings from lower limb muscles to other muscles may need to be further validated.

When comparing the results of the DP-P with the mean + 3 s.d. method, the DP-P method showed a constant delay of about 3 ms in the onset time of both VMO and VL (Fig. 1). Since the delay is consistent in both muscles and across all conditions of testing, this should not affect the comparability of the relative EMG onset time of the two muscles recorded with either method. In the case of studying the absolute EMG onset time, it is highlighted that the findings from the DP-P method cannot be directly compared with the mean + 3 s.d. method.

The difference between the two methods may be explained by the fact that the DP-P and mean + 3 s.d. methods have inherent difference in their constructs. The mean + 3 s.d. method relies on signal processing using a 50 Hz low-pass filter and full-wave rectifier to attenuate the raw EMG signal, so as to transform the AC signals to DC. These processing techniques would smoothen the signals substantially. Since the muscle onset time is the first instance when the signal exceeded the preset threshold, the relatively smoother waveform would have relatively lower threshold as it has proportionally smaller value of mean and standard deviation. In contrast, the DP-P method does not process the raw EMG signal, it only sets the threshold entirely based on the resting signal amplitude, thus the DP-P would take relatively longer time to reach the onset point than the mean + 3 s.d.

This study showed the DP-P method is a reliable way to detect the EMG onset of the vasti muscles in isometric knee extension. The within and between-day reliability is particularly high at 50% of MVC. There is a constant delay of about 3 ms with the DP-P detection method as compared to the mean + 3 s.d. method, thus direct comparison of the absolute muscle onset time of these two methods should not be done.

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Appendix 9

Quantitative assessment for force-displacement relationship of patellofemoral joint

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INTRODUCTION

Patellar gliding has been widely used by therapists and physicians for detecting the patellofemoral joint instability, which is common among persons with anterior knee pain [1]. However, the test is semi-quantitative and may have limited value if being applied in the quantitative research. This study was to evaluate a new quantitative measurement of medio-lateral patellofemoral joint stability.

MATERIALS AND METHODS

13 asymptomatic volunteers (26 knees) participated in the patellar glide test twice with 7 days apart. A DC magnetic motion tracker with 1mm accuracy (pciBIRD, Ascension Technology Corp., Burlington, VT, USA) was used. It consisted of three components, namely, a computer digital signal processing card, a magnetic field transmitter and a magnetic field sensor. The tracker was controlled by an IBM compatible personal computer with a Pentium III-500Hz microprocessor (Intel, Santa Clara, CA, USA) running Windows NT 2000 operating system (Microsoft, Redmond, WA, USA). Additional software used was the Shared Input Devices Controller (Swiss Federal Institute of Technology, Lausanne, Swiss) for collecting 3-axis ordinate data while the sensor was fixed on the manual patellar glide apparatus (Push-pull gage, Aikon Engineering Co., Osaka, Japan) that seated onto the linear bearing (FBW3590-400L, THK Co., Tokyo, Japan) and was supported by the Tri-pod stand (YH-26, Yoga electronics Co., Taipei, Taiwan) (fig 1). The transmitter was located lateral to the tested knee with the subject lying supine. Since only the medial to lateral patellar motion was of interest in this study, data of this axis and were sampled at a rate of 30Hz. With the subject's knees extended and feet together, the ankles were secured and no active movement was allowed during the test, the patellar glide apparatus fitted with the magnetic motion tracker was applied to manually glide the subject's patella laterally in a consistent manner. The gliding force was 0.75 and 1.5 kg respectively, thus the linear displacement of the patella was collected for post acquisition analysis. Both knees were tested separately with three continuous passive gliding in a set for each knee. The average of the three gliding movements was calculated to minimize random error. In order to maintain the subject's vastus medialis (VM) muscle relaxed during the measurement, the mechanomyography (MMG) signal was detected via the air coupled transducer (Pulse wave pickup, 21051D, Hewlett Packard, Palo Alto, CA, USA) that connected to the biofeedback unit (Myotrac, Thoughttechnology, Montreal, Quebec, Canada) for providing visual and audio alarm once the VM's MMG signal in RMS over the preset threshold of 1 μV (fig 2). To establish between-day repeatability, the tests were repeated at 7 days later for each subject and the data were analyzed with ICC.

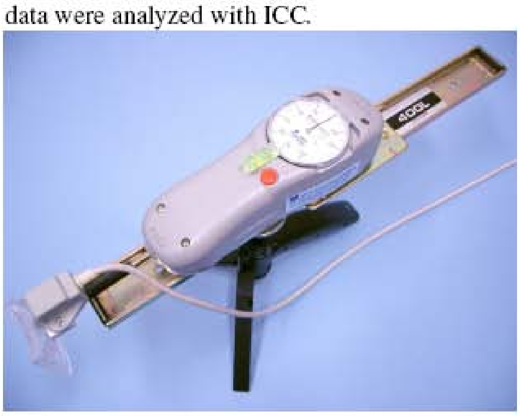

Fig 1: Patellar glide apparatus

Fig 2: Lateral patellar glide test

RESULTS

The ICC (absolute agreement, two-way mixed, CI 95%) value of the measurements was 0.967 for the 0.75 kg lateral displacing force and 0.977 for the1.5 kg force. These ICC values represented excellent between-day repeatability of the measurements.

DISCUSSION AND CONCLUDIONS

Medial patellofemoral ligament (MPFL), which links from the medial epicondyle of the femur to the medical border of the patella, is the primary static stabilizer that resists lateral translation of the patella [2]. The MPFL has been reported to be responsible for 53% of the resistance to the lateral translation at full knee extension; medial patellomensical ligament also contributed 22% of the resistance to the lateral displacing force [3]. Sallay et al [4] showed that 15 out of 16 patents who suffered from acute patellar dislocation had tears of the MPFL from the femoral origin. Also, Garth et al [5] reported that the MPFL was ruptured in all 20 patients who had patellar dislocation, with 10 having disrupted at the margin of the patella and 10 at the adductor tubercle; but pathological laxity was not identified within the medial patellomensical ligament. Therefore, when the lateral patellar displacement under a given force is observed in full knee extension or initial knee flexion while the quadriceps is relaxed, the primary responsible connective tissue is the MPFL. Thus, the equipment evaluated in this study can be used to quantify the lateral patellar passive range of motion and the laxity of the MPFL. This would be clinically important for screening persons whom are prone to patellar dislocation or subluxation [6, 7], thus preventive means such as using protective knee brace or prescribing conditioning exercises to the persons at risk. Also the quantitative measurement can be used to reveal the effectiveness of surgery that reconstructs the MPFL after the patellar dislocation.

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Appendix 10

CORRELATION BETWEEN MRI AND MECHANICAL MEASUREMENTS FOR PATELLAR STABILITY

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INTRODUCTION

Patellar instability is a common cause of patellofemoral pain. While the factors of patellar instability can be examined via mechanical and radiographic means, there is little documentation on the correlation between the imaging and mechanical findings. This study aimed to (I) develop and apply a mechanical method for quantifying passive patellar gliding, (II) record the axial and sagittal views of the knee with magnetic resonance imaging, and (III) correlate the results of I and II.

METHODS

(I) Seventeen able-bodied volunteers (13 males, 4 females, aged 18-35 years) participated in the instrumented lateral patellar gliding test. A DC magnetic motion tracker with 1mm accuracy (pciBIRD, Ascension Technology Corp., Burlington, VT, USA) was used for this assessment. It consisted of a computer digital signal processing card, a magnetic field transmitter and a magnetic field sensor. The sensor was fixed on a patellar glide apparatus (Push-pull gage, Aikoh Engineering Co., Osaka, Japan) which was connected to a plastic linear slide and supported by a plastic lab jack. The transmitter was placed lateral to the tested knee with the subjects lying supine on a wooden bed. A laser-cross projector was located superior to the knees and emitted a cross mark on the bed for standardizing the position of subjects. With the subject's knees extended and feet together, the ankles and lower legs were secured. The patellar glide apparatus was applied manually to glide the patella laterally in a consistent manner. The gliding force was 14.7N and both knees were tested separately with three continuous passive gliding movements. The average of the three movements was calculated to minimize random error as the linear displacement of the patella was collected for off-line analysis. In order to ensure relaxation of the vastus medialis (VM) muscle during the measurement, mechanomyogram (MMG) of VM was monitored via an air-coupled transducer (Pulse wave pickup, 21051D, Hewlett Packard, Palo Alto, CA, USA) and connected to a biofeedback unit (Myotrac, Thought Technology Ltd., Montreal, Quebec, Canada) which provided audio alarm once the MMG signal was over the preset threshold (fig 1).

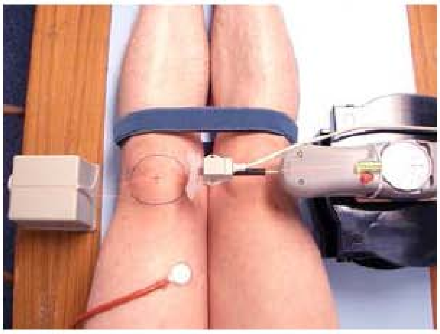

Figure 1: Lateral patellar gliding test

(II) Both knees of the subjects were examined with MRI (Magneton Avanto, Siemens AG, Erlangen, Germany) for sagittal view (T1, TI: 600ms, TR: 1180ms, TE: 4.32ms, slice thickness: 2mm) and axial view (T1, TR: 420ms, TE: 50ms, slice thickness: 3mm) in supine position with knees extended. The MR images were analyzed for the patellar tendon length/patellar length (TL/PL) ratio (Shabshin et al, 2004) and the lateral trochlear inclination (Yannick, et al 2000) (fig 2, 3).

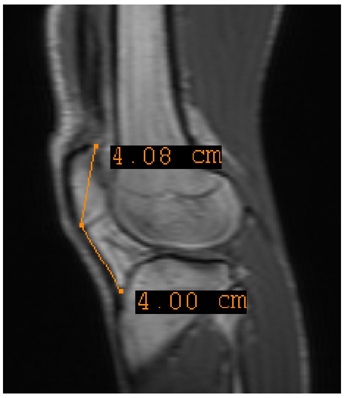

Figure 2: TL/PL ratio

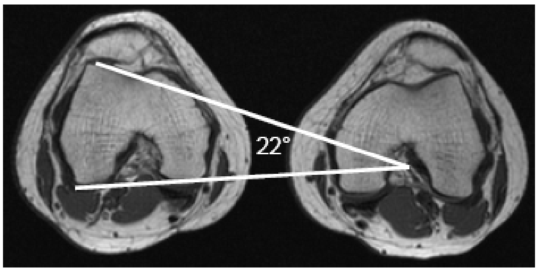

Figure 3: Lateral trochlear inclination

(III) Spearman's correlation coefficient was calculated to examine the relationships between the patellar gliding test and the radiographic measurements.

RESULTS AND DISCUSSION

The results of patellar gliding were between 10.4-23.3mm. All subjects' TL/PL ratio was in the range of 0.74-1.33, and their lateral trochlear inclinations were regarded as normal (13.5-25°). The Spearman's correlation coefficient between the lateral patellar gliding test and the TL/PL ratio was 0.125. The value between the lateral patellar gliding test and the lateral trochlear inclination was -0.458. These implied that the former has a weak association; and the latter has a moderate correlation. These results suggested that the lateral passive mobility of patella may not be related to the length of patellar tendon in terms of the TL/PL ratio among healthy subjects, but may associate with the shape of femoral trochlea in terms of the lateral trochlear inclination. As the angle of the lateral femoral trochlea appeared to be smaller, the patellar stability could be relatively compromised.

SUMMARY/CONCLUSIONS

With the test on able-bodied subjects, the present findings seemed to suggest that the lateral trochlear inclination is a more sensitive index to reflect the lateral patellar instability than the TL/PL ratio.

The instrumented patellar gliding test was moderately correlated to the lateral trochlear inclination measured on MR images. As the gliding test is an inexpensive test for passive patellar stability when compared with the radiographic methods. Physicians may consider the quantitative patellar gliding as the first screening test for patellar disorders.

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Disclosures:
Part of this article was presented in the 2006 American Society of Biomechanics annual meeting, Blacksburg, Virginia.

0894-9115/08/8702-0134/0
American Journal of Physical Medicine & Rehabilitation

DOI: 10.1097/PHM.0b013e31815b42b9

BRIEF REPORT

The Relationships Between the Geometrical Features of the Patellofemoral Joint and Patellar Mobility in Able-Bodied Subjects

ABSTRACT

Wong Y, Ng GYF: The relationships between the geometrical features of the patellofemoral joint and patellar mobility in able-bodied subjects. Am J Phys Med Rehabil 2008;87:134–138.

This study examined the correlations between patellar mobility and geometry of the patella and femoral condyle. Using a custom designed patellar gliding instrument, the lateral mobility of patella in 17 able-bodied subjects was measured and these measurements were compared with the relative patellar tendon length and the inclination of lateral femoral groove obtained from conventional MR imaging. The data showed that there was significant correlation between patellar mobility and inclination of lateral femoral groove ($r = -0.671$, $P = 0.01$). Because hip rotation angle would alter the inclination of lateral femoral groove, clinicians who perform patellar gliding test should be mindful that the hip position is critical for the accuracy of the test.

Key Words: Patellofemoral Joint, MRI, Stability, Patellar Gliding

Patellar instability is a consequence of patellar subluxation or dislocation.[1] If not managed properly, it will predispose the subject to develop patellofemoral pain.[2] Instability of the patellofemoral joint can be defined as the patella being inherently unstable or unable to maintain normal tracking or equilibrium in response to external perturbations.[3] When clinicians treat people with patellar instability, knowing the magnitude and cause of the problem is essential for the planning of the programs and evaluation of the treatment progress.[4] Clinically, physicians or therapists can measure patients' *Q* angle for estimating the lateral patellar force.[5] Different versions of patellar tilt and gliding test have been established for assessing the resting position and passive mobility of the patella.[6–8] Radiographically, there are various protocols developed for the imaging of the patellofemoral joint.[9] Whereas the magnitude and cause of patellar instability can be examined via mechanical and radiographic means, there is little information on the correlations between the mechanical and radiographic findings. This study was therefore conducted with able-bodied subjects to (1) develop and apply a mechanical method for quantifying passive patellar gliding, (2) record the axial and sagittal views of the knee with magnetic resonance imaging (MRI), and (3) correlate the results of 1 and 2. If the method in this study was found to be feasible, it could be developed into a clinical assessment for subjects with knee pathologies.

METHODS

Subjects

Seventeen able-bodied subjects (13 males, 4 females) aged between 18 and 35 yrs (mean ± SD = 26.7 ± 5.1) participated in the study on voluntary basis. They had no history of patellar subluxation/dislocation, or operations to their lower limbs. The study was approved by the human ethics subcommittee of the Hong Kong Polytechnic University before data collection.

Instrumented Lateral Patellar Gliding Test

With the subjects lying supine on a wooden plinth, knees extended and feet together, the ankles and lower legs were strapped to the plinth so as to restrict any active movement. A laser-cross projector (SL01, Land, Hong Kong) was mounted on a jig above the knees which emitted a cross mark on the plinth for standardizing the position of the subjects. A patellar gliding apparatus comprising an assembly with a force gauge transducer (AEF-5, Aikon Engineering Co., Osaka, Japan), an electromagnetic motion tracker (pciBird, Ascension Technology Corp., Burlington, VT), a plastic linear bearing, and an adjustable platform (Poly-Jaque plastic lab jack, Scienceware, Pequannock, NJ) was used to manually glide the subject's patella laterally under a constant velocity of approximately 2 cm/sec (Fig. 1). The gliding force was 14.7 N (1.5 kg), and the mediolateral displacement of the patella was collected for post acquisition analysis with the Excel software (Microsoft, Redmond, WA).

Both knees were tested separately with three continuous passive gliding movements and the average of the three recordings was calculated to minimize the random error. To help the subjects relax their quadriceps muscles during the measurement, biofeedback signal was provided with mechanomyography (MMG) of the quadriceps muscles. The MMG signal was detected via an air coupled transducer (Pulse wave pickup, 21051D, Hewlett Packard, Palo Alto, CA) connected to a biofeedback unit (Myotrac, Thought Technology, Montreal, Quebec, Canada) which would provide an audio alarm when the quadriceps' MMG signal (root mean square, 100-ms time constant) was over a preset threshold of 1 μV (Fig. 1).[10] A pilot study with 13 other subjects (26 knees from 10 males and 3 females) was completed to reveal intrarater reliability for the measurement with a 7-day interval and the intraclass correlation coefficient (consistency, two-way mixed) value was 0.96 (95% CI: 0.94–0.98), which indicated very high reliability of the measurement.

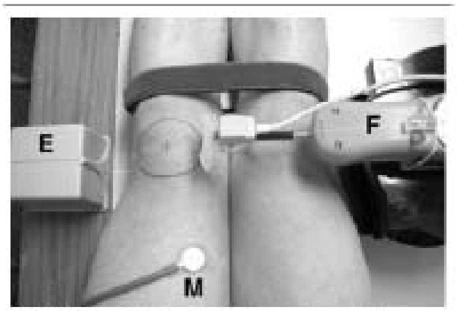

FIGURE 1 *Setting of passive patellar mobility measurement: (E) electromagnetic tracker, (M) mechanomyographic biofeedback, (F) force gauge transducer.*

Magnetic Resonance Imaging (MRI)

After the patellar gliding test, the subjects received MRI of their knees and lower thighs. A 1.5-T MRI scanner with surface coils (Magneton Avanto, Siemens AG, Malvern, PA) was used to record the sagittal (T1, TI: 600 msecs, TR: 1180 msecs, TE: 4.32 msecs, slice thickness: 2 mm) and axial views (T1, TR: 420 msecs, TE: 50 msecs, slice thickness: 3 mm) of the subjects' knees and lower thighs with each subject in the same positioning as the patellar gliding test. Foam blocks were used to support the surface coils to prevent the anterior knees from being compressed by the coils. Image datasets were saved in DICOM format for offline analysis.

Analyses were done with an image analysis software (MIPAV, version 2.0, National Institutes of Health, Bethesda, MA) and a graphic digitizer (CTE-430, Wacom, Vancouver, WA). The MR images were analyzed for the patellar tendon length/patellar length (TL/PL) ratio,[11] the lateral femoral trochlear inclination[12] and the biomechanical inclination angle (Fig. 2). Their interrater reliability of these measurements were tested previously with the ICC values of between 0.98 and 0.99.[11,12]

Data Analysis

Because both legs of each subject were measured, this resulted in 34 data. Pearson's coefficient ($P = 0.05$, two tailed) was calculated to examine the relationships between the patellar gliding test and the MRI measurements.

RESULTS

Table 1 presented the results of the descriptive data of the patellar gliding test and MRI measurements. The Pearson's coefficient of correlation between the patellar gliding test and the TL/PL ratio

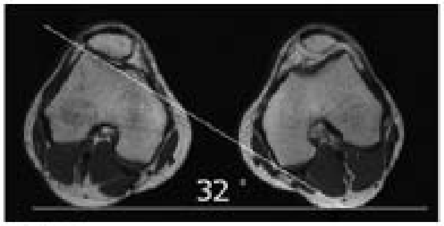

FIGURE 2 *Biomechanical inclination angle was subtended by a line tangential to the lateral trochlear facet and the supporting surface on the first craniocaudal image that demonstrated cartilaginous trochlea.*

was 0.312 ($P = 0.073$). The correlations between patellar gliding and the lateral femoral trochlear inclination and the biomechanical inclination angle were −0.535 ($P = 0.001$) and −0.671 ($P = 0.001$), respectively (Fig. 3).

DISCUSSION

We measured the lateral patellar mobility and correlated it with the geometrical parameters of TL/PL ratio, lateral trochlear inclination, and biomechanical inclination angle in consideration of the high reliability of these measurements.[11,12] The femoral sulcus angle was not used as it was reported to be nonreproducible by Koskinen and Kujala[13] and Koskinen et al.[14] Shabshin et al.[11] studied 245 knees with MRI and reported that a TL/PL ratio of between 0.74 and 1.5 to be normal. If the ratio is >1.5, it is classified as patellar alta and this implies a potentially unstable patella due to the excessively long patellar tendon. With a high sensitivity of 91% and specificity of 81%, Carrillon et al.[12] postulated that at fewer than 11 degrees of lateral trochlear inclinication, it was a diagnostic determinant of patellar instability. According to these criteria, all the subjects in this study would be regarded as within the norm.

TABLE 1 Descriptive data of patellar gliding test and MRI-based measurements

Parameter	n	Mean	SD	Range
Patellar gliding test, mm	34	14.2	4	10–25
TL/PL ratio	34	0.95	0.14	0.74–1.33
Lateral trochlear inclination, degrees	34	19.4	3.9	14–27
Biomechanical inclination angle, degrees	34	23.1	3.5	14–28

Our findings suggested that the lateral passive mobility of patella was weakly correlated to the length of patellar tendon (TL/PL ratio), but it was significantly correlated to the geometry of the femoral trochlea in terms of the lateral trochlear inclination and the biomechanical inclination angle. As these angles became smaller, the passive patellar mobility tended to increase. The values of correlation suggested that the biomechanical inclination angle to be more predictive of the passive patellar mobility than the lateral trochlear inclination.

The lateral trochlear inclination would not be affected by the hip rotation angle or the angle of femoral torsion,[15] but the biomechanical inclination angle would be affected by these angles. This point is clinically important because when a clinician performs the patellar gliding test (manual or instrumented) on a patient, the hip rotation angle (limb position) must be controlled as it can affect the biomechanical inclination angle that may alter the magnitude of the lateral patellar displacement.

We believed that when the patella was glided laterally, the gliding force was directed horizontally and diagonally downward due to the irregular shape of the patella. The force would induce a reaction force on the lateral facet of the femoral trochlea, and it was this reaction force that generated friction in the lateral patellofemoral joint which reduced the patellar translation. An increase in the inclination angle would therefore hinder the lateral patellar gliding movement while a decrease of that angle would facilitate the gliding movement.

The patellar gliding test was only performed for the lateral side since majority of patellar disorders happen on the lateral side.[1] Fulkerson[16] and Nonweiler and DeLee[17] confirmed that medial patellar subluxation is clinically uncommon and in the rare cases of medial subluxation, they are usually the complication after excessive lateral retinacular release.

Fithian et al.[8] were the first group to use instrumented patellar gliding measurement and were able to distinguish the patellar force–displacement relationship between subjects with and without patellar dislocation. They tested the knees at 30 degrees of flexion, whereas in the present study, a fully extended knee position was used because the patella would be minimally constrained at full extension. When the knee is flexed to more than 20 degrees, the patella would be drawn into the trochlear sulcus due to the closing angle between the quadriceps and patellar tendon in the sagittal plane, hence the patellar stability and the patellofemoral joint contact area would increase.[18] It has been suggested that the lateral gliding test done with an extended knee can test the medial patellofemoral ligament which is a pri-

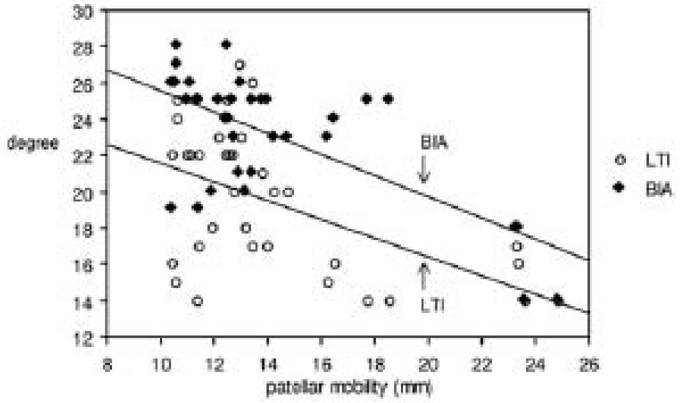

FIGURE 3 *Bivariate plot with linear regression lines showed relationships between patellar mobility and MRI-based measurements. LTI, lateral trochlear inclination; BIA, biomechanical inclination angle.*

mary medial patellar restraint susceptible to injury during patellar dislocation.[19] Also, the corresponding gliding force would need to be increased with the knee flexion angle for a given displacement. The lower gliding force of 3.3 lbs (1.5 kg) in the current study as opposed to 5 lbs in the study by Fithian et al.,[8] was a precautionary measure in case of a participant who might have an inherently unstable patella.

There are some limitations in this study that need to be addressed. Since the subjects being tested were all young and asymptomatic, the findings may not be directly generalized to older subjects or those with knee pathologies. In the patellar gliding test, we used a standard gliding force of 14.7 N rather than a percentage of the subjects' body weight, the potential effect of the nonindividualized gliding force on the correlations was therefore unknown. We only tested 17 subjects (34 knees) with most of them being males ($n = 13$). The potential gender difference in lower limb alignments or soft tissue compliance may affect the femoral geometry–patellar mobility correlations because it has been reported that females have a relatively higher TL/PL ratio.[11] Stefancin and Parker[20] report that the male to female ratio was 46:54 in subjects with patellar dislocation for the first time, but females have significantly higher risk of recurrent dislocation. Therefore, further study with more female participants and people with patellar disorders is warranted to confirm the current findings.

CONCLUSION

We conclude that the inclination of lateral femoral groove is a significant factor affecting the lateral patellar mobility rather than the patellar tendon length in able-bodied subjects. Because the hip position can change the biomechanical inclination angle, clinicians should standardize the hip position when performing the lateral patellar gliding test.

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Appendix 12

RELIABILITY OF HAND-FREE ULTRASOUND MEASUREMENT FOR VASTUS MEDIALIS OBLIQUUS

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INTRODUCTION

Atrophy of the vastus medialis obliquus (VMO) muscle is relatively common among patients with patellofemoral pain syndrome. In order to monitor the effectiveness of rehabilitation program for strengthening the muscle, clinicians may measure the distal circumference of the thigh with tape. However, the distal thigh contains VMO as well as other muscles, the circumference measurement may not precisely address the change for the VMO muscle.

B-mode ultrasound can visually quantify muscular cross-sectional area, and is usually controlled by the operator manually holding the ultrasound probe. However, little has been done to reveal the reliability of such measurement for the VMO. Electromagnetic tracking (e.g. pciBird, Ascension, USA) and 3D digitizing (e.g. Microscribe, Immersion, USA) instruments have been used to ensure accurate positioning of the ultrasound probe at the same muscle for periodical scanning. Despite the costs of these positioning devices, the accuracy of the electromagnetic tracker is susceptible to adjacent metal (e.g. hospital bed) and a lack of mobility of the 3D digitizer is not very compatible to clinical settings.

The present study aimed (1) to design a hand-free ultrasound system with economic and portable manner; (2) to examine the reliability of the system for measuring the VMO cross-sectional area by comparing it with a gold standard and between-day measurements.

METHODS

Design: A custom-made mechanical stand adjustable in height and tilt angle was used to statically hold an ultrasound probe (Voyager, Ardent Sound, USA) (fig 1).

Figure 1: Ultrasound probe holder

A laser pointer projected a cross mark at the suprapatellar level and a sliding ruler was used to measure the distance between the sole of feet and the suprapatellar border (fig 2). The distance and laser-cross acted as landmarks for the repeatable positioning of the ultrasound probe while a subject supine lying with knee extended and feet secured.

Figure 2: Cross-section locator

An ultrasound gel pad (Aquaflex, Parker Lab, USA) was molded to match the shape of the ultrasound probe and the VMO underneath. The pad weighed about 15g, thus it would not deform the VMO significantly (fig 3).

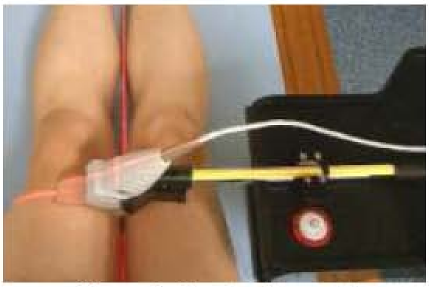

Figure 3: B-mode ultrasound

Testing procedures: Twelve VMO muscles (6 healthy volunteers, 4 males & 2 females) at suprapatellar level were imaged using the B-mode ultrasound and MRI (Magneton Avanto, Siemens AG, Germany) within 3 hours. Both set of the images were saved in DICOM format for offline comparison of the cross-sectional area of VMO (fig 4).

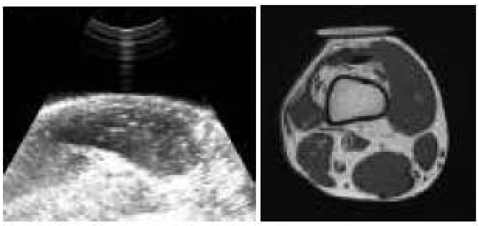

Figure 4: VMO in ultrasound and corresponding MRI image

The twelve VMO muscles were ultrasonically scanned at one week later. Intraclass correlation coefficient ($p=0.05$) was calculated for the consistency of the two sets of the VMO cross-sectional area.

RESULTS

The mean difference was 5.7% (SD=3%) between the VMO measurement based on the ultrasound and MRI images (12 knees). The ultrasound method slightly underestimated the muscle size compared with the MRI.

The ICC result for the 7-day interval for the VMO ultrasound measurement (12 knees was 0.96 (95% CI=0.89-0.99).

DISCUSSION

Results of this study reveal that the hand-free ultrasound measurement to be comparable with MRI measurements and the system has satisfactory between-day repeatability for the VMO muscle. The application should not be limited to the VMO; it is potentially applicable to other skeletal muscles for diagnostic or biofeedback training purpose.