計畫編號：NHRI-EX91-9126EP

國家衛生研究院九十一年度整合性醫藥衛生科技研究計畫

前十字韌帶損傷及重建後膝關節之生物力學與神經肌肉適應之分析

Biomechanics and Neuromuscluar Adaptation in ACL patients

年度成果報告

執 行 機 構：中國醫藥學院

計 畫 主 持 人：許弘昌

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計畫名稱：

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執行機構：中國醫藥學院、台灣大學、高雄醫學大學

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關鍵字：前十字韌帶、生物力學、電腦模型、動作分析、運動功能、肌力、

自主徵召、動作控制

壹、九十一年度計畫研究成果摘要

膝關節由脛骨與股骨和周圍組織如韌帶等所構成，其關節接觸面與韌帶共同決定了在被動活動時的運動情形，再加上神經支配的周圍肌肉提供主動支持，而共同控制膝關節的動作。正常的關節使人類完成許多日常生活的能力，包括行走位移，站或坐等等。前十字韌帶不但提供膝關節穩定作用之外，也傳回本體感覺動作及受力的訊息以作爲肌肉動態控制膝關節的依據。若前十字韌帶受損，將影響正常動作和肌肉控制，使病人從事活動時，會有不適的情況。本計畫全面性地探討前十字韌帶在膝關節功能中所扮演的角色，包括建立其生物力學模型及觀察病患在損傷之後的動作以及肌肉表現的改變，分別於本計畫的三個子計畫中探討。依序說明如下。

第一子計畫之目的在建立三維膝關節電腦模型，應用於靜態或活動，以了解前十字韌帶對控制膝關節活動度和穩定度所扮演的角色，也希望對前十字韌帶的重建手術和治療有所幫助。本階段膝關節模型主要用以模擬活動度、穩定度與等長肌力等測試。活動度測試之模擬係在肌肉未收縮的情形下，由關節面以及韌帶之幾何形狀，依最小能量法計算關節活動範圍、肌肉力臂與作用線、關節面接觸型態等資料。關節穩定度測試之模擬則是在選定膝關節彎曲角度且肌肉未收縮的情形下，施外力於脛骨上，算出脛

骨之平移量以代表關節之鬆弛度。等長肌力測試之模擬則考慮脛骨受一限制力而單一肌肉做最大收縮時，決定各元件受力情形。

利用 MRI 及 CT 影像資料重建膝關節幾何形狀，包括股骨、脛骨、髕骨、各主要肌群及韌帶。骨骼與關節面初步假設爲剛性。關節面間之接觸型態取決於關節角度、外力、各關節面之曲率以及肌肉與韌帶受力與變形情形。膝關節韌帶包括十字韌帶及內外側韌帶則模擬爲彈性的纖維束。影響膝關節的肌肉則根據 Hill's equation 模擬爲具主動收縮能力之單元，考慮肌肉長度、速度、激化程度與所產生肌力之關係。相關肌肉參數如最佳肌纖維長度、生理截面積等取自於文獻資料，依受試者肌力實驗資料調整之。

電腦模型之個人化、驗証與正常膝關節於單關節測試動作時之力學分析等，透過第二子計劃實驗量測配合進行並驗証。包括利用三維動作量測系統之紅外線攝影機，攝取肢段在空間中的位置以同步擷取，以 arthrometer (KT2000)量取關節鬆弛度，並與第三子計劃整合設計並製作一個等長肌力測試裝置，以測量膝關節肌肉等長收縮時作用於脛骨之限制力量。

MRI、CT 資料及反光標記資料爲定義膝關節模型幾何形狀之個人化。鬆弛度實驗所得資料用以調整膝關節模型之材料特性，特別是韌帶纖維之彈性係數及初始長度。最大等長肌力測試資料則用以調整肌肉模型參數，使整個膝關節生物力學模型得以充份反應受測者膝關節的特性。個人化之

後，即模擬活動度試驗，所得脛股關節屈曲角度與另外兩個平面角度具一對一關係，証明該關節係具有一個自由度，與文獻一致(Wilson et al., 1998)。鬆弛度測試以及最大等長肌力測試模擬結果與實際量得數據均相當一致。

第二子計畫之目的在於利用三度空間立體攝影術之進行動作分析，以探討前十字韌帶損傷患者因受傷所造成其運動功能之變化，以及其後接受前十字韌帶重建手術後的短、長期效果。第一年度的計劃，研究內容著重於建立臨床患者與研究室的轉介、病患追蹤模式、研究人員訓練、整合應用完整的三度空間動作分析儀，建立動作分析研究之相關研究流程、收集正常受試者與前十字韌帶損傷患者之從事不同功能性活動時的動作特徵。

第二子計畫今年度已建立互動良好的的病患轉介模式，自中國醫藥學院附設醫院骨科收集前十字韌帶斷裂的病患。截至目前爲止，已收集單純的單側前十字韌帶斷裂病人十名，爲第一組；曾接受前十字韌帶自體髕骨韌帶重建術之病患十名，爲第二組；另外並徵求十位正常受試者，作爲控制比較組，爲第三組。實驗進行先收集每位受試者之基本描述性資料，膝關節鬆弛度測試(KT-2000)，膝關節運動功能問卷(Lysholm score)，動作分析部分共進行四項活動，包含平地行走、跨越障礙物、由坐到站、上下樓梯等，並研究穿上膝關節支架之後平地行走與上下樓梯的動作型態變化。

研究結果顯示，膝關節支架對平地行走與上下樓梯的動作型態並無顯

著立即性的影響，但因今年度的研究爲一橫斷面研究，其長期性的效果上有待未來的追蹤與研究。在跨越障礙物部分，發現在正常受試者的膝關節與髖關節屈曲角度會隨障礙物高度增加而增加，然而支撐腳的力矩(moment)卻沒有隨之變化；而前十字韌帶受損患者，受傷側支撐時的最大力矩值會下降，顯示減少膝關節伸直肌力矩的意圖。在從事由坐到站時，同樣的隨不同高度，主要只有關節屈曲角度的變化，在力矩方面的變化並不明顯，而與其他功能性活動相較之下，從事由坐到站下肢須承受較大的冠狀面和橫斷面上的力矩，因而需要較多的外展與外轉的肌群作用已完成此項活動。而在上下樓梯的部分，穿著膝關節支架同樣對於活動並沒有顯著的影響，而在上樓梯時，患側的膝關節彎曲角度顯著減小，而無論是上樓梯或下樓梯時，膝關節與髖關節伸直肌的力矩都顯著的下降。

初步結果顯示，前十字韌帶重建後的患者，膝關節被動鬆弛度有所改善，因而提供膝關節靜態的穩定性，但是在受傷後與手術後的運動功能仍有不同於正常人的表現，其中的神經肌肉適應的調整而影響膝關節生物力學的機制尚不能完全由今年度的結果解釋。文獻中膝關節穩定性的評估多侷限在靜態或是半靜態的狀況，然而動態活動時，膝關節周圍的被動與主動限制的組織，包含肌肉與韌帶，其受力並不容易由初步的評估結果得知，因而本計劃後續的追蹤並整合將結合其他子計劃的研究成果，希望能初步

解決上述的問題，而可以對整個動作與功能恢復機轉有整體性的瞭解。

第三子計畫之目的在於探討前十字韌帶損傷患者其神經肌肉系統的功能變化與其後接受前十字韌帶重建手術後的短、長期效果。本年度研究內容除著重於建立肌力研究室及建立交流機制之外，尙研究正常受試者與前十字韌帶損傷患者之膝關節伸／屈肌之肌力與動作特徵。

此子計劃今年度的實驗可分成兩大部分，首先是針對股後肌群(hamstrings)自主徵召能力的測試。對於前十字韌帶損傷患者而言，膝關節前後肌肉的平衡與協調十分重要，因此股四頭肌與股後肌群的運動單元(motor units)是否能被自如的運用、徵召，是此子計劃本年度研究的重點。在一般健康者身上，包括股四頭肌等的許多肌肉都已被證實能夠做出所謂的『最大自主收縮』，但或許因爲實驗技術上的困難，至今尙未有相關研究。爲了對前十字韌帶損傷患者在重建手術前後的肌力變化機制能夠有更正確的瞭解，本研究嘗試利用『抽搐上加』(twitch superimposition)技術探討股後肌群的自主徵召性質。結果發現，在健康的股後肌群做最大自主收縮時，的確可以做到完全徵召，而在不同程度的自主收縮下，可找出一個自主收縮力量-徵召的曲線(voluntary activation curve)；利用此不但可以判斷此肌肉的收縮能力，也可以用於推測股後肌群在正常收縮狀況下所應產生的肌力。

另一研究重點爲前十字韌帶損傷與重建患者之肌力與動作特徵。除術

前的評估，患者需於術後三及六個月再回診評估。至今共有五位患者符合受試者資格，其中三位已完成第二次(術後三個月)評估。另外本年度徵召15 位健康志願者做爲對照比較。初步結果顯示，術前患者之患側股四頭肌與股後肌群力量均有下降，尤以做等長及慢速等速收縮時較爲明顯。另外，患者兩側肌肉都有明顯的自主徵召困難；患側比健側、股四頭肌比股後肌群明顯。而在已做過第二次評估的三位患者身上發現，患側股四頭肌除在術前即有明顯的肌無力，術後此現象更顯著，特別是在膝關節彎曲 10-20°時。而患側股後肌群在術前、術後三個月並沒有明顯的變化。在健側，不論是股四頭肌或是股後肌群，肌力與自主徵召的能力都比之前的表現要好。

第三子計劃目前所呈現出來的結果證實了之前的推測，也就是前十字韌帶損傷與重建患者的膝關節肌肉功能表現除了肌力的變化外，自主徵召能力也呈現了相對應的改變，也可能是造成肌肉無力的原因之一。這些改變除了可以作爲術後恢復的指標外，尚可以綜合下一年度有關神經反射的研究以對整個動作控制機轉有一整體性的瞭解，並可用以驗證第一子計劃的電腦化膝關節模型，以及整合應用第二子計劃所測得的運動功能表現。

本計劃的三個子計劃在第一年中均依照計劃之預定進度，也已充分顯示出各子計劃間相互合作的必要。隨著各子計劃累積的研究成果，相信在未來一年將能夠發揮更多的整合與成果。

貳、九十一年度計畫著作一覽表

註：群體計畫(PPG)者，不論是否提出各子計畫資料，都必須提出總計畫整合之資料

若為群體計畫，請勾選本表屬於：☐子計畫；　或　☐總計畫(請自行整合)

1.列出貴計畫於本年度中之所有計畫產出於下表，包含已發表或已被接受發表之文獻、已取得或被接受之專利、擬投稿之手稿（manuscript）以及專著等。

2.「計畫產出名稱」欄位：請依「臺灣醫誌」參考文獻方式撰寫；

3.「產出型式」欄位：填寫該產出為國內期刊、國外期刊、專利、手稿或專著等。

4.「SCI」欄位：Science Citation Index，若發表之期刊為 SCI 所包含者，請在欄位上填寫該期刊當年度之 impact factor。

5.「致謝與否」欄位：若該成果產出有註明國家衛生研究院委託資助字樣者，請打勾。

序號	計　畫　產　出　名　稱	產出型式	SCI*	致謝與否
1.	Tung-Wu Lu, Hsiu-Chen Lin, Te-Chang Kao, Horng-Chaung Hsu. (2002) Stepping over obstacles during locomotion in anterior cruciate ligament patients. VI World Congress of Biomechanics, Calgary, Alberta, Canada.	國外研討會		✓
2.				
3.				
4.				
5.				
6.				
7.				
8.				

*本表如不敷使用，請自行影印。

參、九十一年度計畫重要研究成果產出統計表

註：群體計畫(PPG)者，不論是否提出各子計畫資料，都必須提出總計畫整合之資料

若爲群體計畫，請勾選本表屬於: □子計畫； 或 □總計畫(請自行整合)

(係指執行九十一年度計畫之所有研究產出成果)

科技論文篇數			技術移轉			技術報告 篇
	國內	國外	類型	經費	項數	技術創新 1 項
期刊論文	篇	篇	技術輸入	千元	項	著作權(核准) 項
研討會論文	1 篇	篇	技術輸出	千元	項	
專著	篇	篇	技術擴散	千元	項	專利權(核准) 項

〔註〕:

期刊論文：指在學術性期刊上刊登之文章，其本文部份一般包含引言、方法、結果、及討論，並且一定有參考文獻部分，未在學術性期刊上刊登之文章（研究報告等）與博士或碩士論文，則不包括在內。

研討會論文：指參加學術性會議所發表之論文，且尚未在學術性期刊上發表者。

專　　著：爲對某項學術進行專門性探討之純學術性作品。

技術報告：指從事某項技術之創新、設計及製程等研究發展活動所獲致的技術性報告且未公開發表者。

技術移轉：指技術由某個單位被另一個單位所擁有的過程。我國目前之技術轉移包括下列三項：一、技術輸入。二、技術輸出。三、技術擴散。

技術輸入：藉僑外投資、與外國技術合作、投資國外高科技事業等方式取得先進之技術引進國內者。

技術輸出：指直接供應國外買主具生產能力之應用技術、設計、顧問服務及專利等。我國技術輸出方包括整廠輸出、對外投資、對外技術合作及顧問服務等四種。

技術擴散：指政府引導式的技術移轉方式，即由財團法人、國營事業或政府研究機構將其開發之技術擴散至民間企業之一種單向移轉（政府移轉民間）。

技術創新：指研究執行中產生的技術，且有詳實技術資料文件者。

肆、九十一年度計畫重要研究成果

註：群體計畫(PPG)者，不論是否提出各子計畫資料，都必須提出總計畫整合之資料
若爲群體計畫，請勾選本表屬於: □子計畫； 或 □總計畫(請自行整合)

1.計畫之新發現或新發明

本計畫研發出的新膝關節模型，已經能合理地模擬膝關節的力學性質，經未來細部的調整後，應可使用於臨床現象的預測與解釋方面。由步態分析中也發現前十字韌帶受損病人的關節力學經重建手術後有力學性質上的改變，但術後病人在動作功能上和正常人還是有差別，對於此動態功能的影響還有待接下來的研究以提供解釋。對於前十字韌帶受傷病人的肌肉適應分析，顯示了許多有趣的結果。例如患側腳的自主徵召困難、及徵召速度減緩等等，待未來探討膝關節附近的反射能力之後，能對整個膝關節肌肉控制的影響作更詳盡的解釋。

2.計畫對學術界或產業界具衝擊性（impact）之研究成果

由步態分析中發現功能性支架對於前十字韌帶損傷病人並無立即性的影響，雖然長期性的效益仍不能看出，未來在對功能性支架長期性的效益作評估後，相信能更進一步對它的價値有所探討。

3.計畫對民眾具教育宣導之研究成果

此部份將爲規劃對一般民眾教育或宣導研究成果之依據，請以淺顯易

懂之文字簡述研究成果，內容以不超過 300 字為原則。

前十字韌帶是膝關節相當重要的結構之一，但不幸地在運動或活動中這條韌帶受傷的機會很高，也因此許多人（特別是運動員）因此而終止了他們的運動生涯，或無法執行一些較具挑戰性的活動。本計畫除了研究大腿及膝關節附近的肌肉與韌帶在接受手術及治療前後會有甚麼樣的改變之外，研究的結果將會使民眾能夠更了解膝蓋的功能表現以及物理治療的效用，這些結果也可以成為進一步復健運動或功能訓練的指標。前十字韌帶受傷後需提早治療，因為可能造成許多後遺症：包括肌力的不足、膝關節提早老化、平衡反應變差等等。不過近年來骨科手術的進步及物理治療計畫的改進，使得前十字韌帶受傷的患者再度能夠重拾膝關節的控制及運動能力，一般的日常生活或運動也不再成為問題。若是能更加了解前十字韌帶受傷後、手術前、手術後等一連串的變化，相信能帶給大眾更好的治療，也能減少不必要的二度傷害。

4.技術移轉（註明成果或技術名稱、移轉對象及概略情形）

5.技術推廣（註明成果或技術名稱、移轉對象及概略情形）

6.業界合作成效（註明成果或技術名稱、移轉對象及概略情形）

7.成效評估（技術面、經濟面、社會面、整合綜效）

8.下年度工作構想及重點之妥適性

本年度三個子計畫已經有初步成果，未來的工作將朝向更進一步的整合與合作，第一子計畫的關節模型將可協助分析解釋第三子計畫的肌肉適應分析結果及第二子計畫的現象。而第二子計畫和第三子計畫也可以提供作爲第一子計畫模型的調整等等。

第一子計畫下年度除將繼續調整膝關節模型外，重點將在死體實驗設備之設計製作與實驗之執行，以及利用實驗資料進一步驗證關節模型，也將用以協助第二與第三子計畫建立膝關節在動態情況下的模型。第二子計畫下年度的重點在追蹤術後病人動作參數的長期各階段變化，並探討功能性支架對病人長時間使用的影響。第三子計畫下一年度的研究主軸在於膝關節肌肉之神經反射與平衡功能的評估。這些研究項目所需之儀器設備尚待採購，不過已經擬好整個配合的方式以節人員的培訓也已經展開。今年度的肌力與自主徵召實驗會在下一年度繼續，以求有明確的研究成果。

9 檢討與展望

本年度三個子計劃分別進行計劃之預定進度，並定期召開內部主持人會議將研究成果加以統合，雖執行機構分別位於北中南三處，但在計劃執行過程中，分別克服計劃執行上的困難與瓶頸，透過密切的聯絡與討論，愈發顯得三個子計劃是緊密的互相關聯且相輔相成。第一子計劃本年度重點在於個人化膝關節模型方法之建立與初步驗證，未來將進一步推廣用於更多受試者與病人，以建立其臨床應用性，並整合至一既有之下肢模型供第二子計劃下之動作分析之用。第二子計畫今年主要的工作在建立起轉介模式，多收集新的病人，所以對於病人的追蹤尚有時間上的限制，下年度重點將調整至病人的長期追蹤上，並整合第三子計劃對於肌肉主動徵召的相關研究方法追蹤病人的恢復情形。第三子計畫本年度最大的限制來自於設備及病患來源，所幸在各單位配合下，仍能有不錯的成果。在未來一年的研究中，為了突破人數的限制，也許需要增加合作的單位或醫師，但由於開刀以及重建方式的選擇、術前術後的物理治療等都有可能影響結果，對此尚須審慎考量。三個子計劃在經過一年的緊密合作，相信未來也能分別進行有所突破，並能整合為一，進而增進學術上與臨床上之成果與貢獻。

伍、九十一年度計畫所培訓之研究人員

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種 類			人數	備 註
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		已結訓		
	2. 碩士級研究人員	訓練中		
		已結訓	1	吳曉臻
	3. 學士級研究人員	訓練中	3	黃于珊、趙偉鈞、蔡伊謹
		已結訓	1	黃筠婷
	4. 其他	訓練中		
		已結訓		
兼任人員	1. 博士班研究生	訓練中	2	林秀真、劉玫舫
		已結訓		
	2. 碩士班研究生	訓練中	4	洪啓峰、高德昌、蔡宗遠、陳淑雅
		已結訓		
	3. 大學部學生	訓練中	9	陳亮伃、王佑予、何思穎、吳俊男、蔡佩怡、冉彩萍、謝劭文、林昆瑩、蔡秉憲

		已結訓	1	魏文一
	4. 其他	訓練中	5	潘慧芬、蕭宜芳、江采曄、李軒西、李俊弘
		已結訓	1	黃志偉
醫　　師		訓練中	2	黃茂雄、周伯禧
		已結訓		
特 殊 訓 練 課 程				

註： 1.特殊訓練課程請於備註欄說明所訓練課程名稱

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陸、參與九十一年度計畫所有人力之職級分析

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職級	所 含 職 級 類 別	參與人次
第一級	研究員、教授、主治醫師	2 人
第二級	副研究員、副教授、總醫師	3 人
第三級	助理研究員、講師、住院醫師	3 人
第四級	研究助理、助教、實習醫師	8 人
第五級	技術人員	8 人
第六級	支援人員	8 人
合計		32 人

〔註〕

第一級：研究員、教授、主治醫師、簡任技正，若非以上職稱則相當於博士滿三年、碩士滿六年、或學士滿九年之研究經驗者。

第二級：副研究員、副教授、助研究員、助教授、總醫師、薦任技正，若非以上職稱則相當於博士、碩士滿三年、學士滿六年以上之研究經驗者。

第三級：助理研究員、講師、住院醫師、技士，若非以上職稱則相當於碩士、或學士滿三年以上之研究經驗者。

第四級：研究助理、助教、實習醫師，若非以上職稱則相當於學士、或專科滿三年以上之研究經驗者。

第五級：指目前在研究人員之監督下從事與研究發展有關之技術性工作，且具備下列資格之一者屬之：具初（國）中、高中（職）、大專以上畢業者，或專科畢業目前從事研究發展，經驗未滿三年者。

第六級：指在研究發展執行部門參與研究發展有關之事務性及雜項工作者，如人事，會計、秘書、事務人員及維修、機電人員等。

柒、參與九十一年度計畫所有人力之學歷分析

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類別	學歷別	參與人次
1	博士	2　人
2	碩士	3　人
3	學士	10　人
4	專科	0　人
5	博士班研究生	2　人
6	碩士班研究生	3　人
7	其他	9　人
合計		29　人

捌、參與九十一年度計畫之所有協同合作之研究室

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機構	研究室名稱	研究室負責人
台灣大學醫學工程學研究所	骨科工程暨動作分析實驗室	呂東武
中國醫藥學院物理治療學系	動作分析實驗室 運動生物力學實驗室	許弘昌
高雄醫學大學附設紀念醫院復健科	等速肌力評估室	黃茂雄 主任

玖、九十一年度之著作抽印本或手稿

依「貳、九十一年度計畫著作一覽表」所列順序附上文獻抽印本或手稿。

STEPPING OVER OBSTACLES DURING LOCOMOTION IN ANTERIOR CRUCIATE LIGAMENT PATIENTS

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INTRODUCTION

Stepping over obstacles is a common activity of daily living and frequently caused fall by tripping over [1]. A safe and successful obstacle-crossing requires stability of the stance limb and sufficient foot clearance in the leading limb. These requirements may not be guaranteed in patients with anterior cruciate ligament deficiency (ACLD) that impairs both the structural stability and sensory feedback of the joint [2]. The influence of these impairments on the performance in stepping over obstacles during locomotion has not been reported in the literature. The present study addresses this issue by comparing the foot clearances and knee joint kinetics in ACLD and ACL-reconstructed (ACLR) patients. The advantage of reconstruction of ACL in this regard is discussed.

MATERIALS AND METHODS

Three groups of subjects participated in this study: six unilateral ACLD subjects (age: 23.2±4.9 years, height: 171.7±5.8 cm, weight: 70.8±7.1 kg), seven unilateral ACLR subjects (age: 27±8.1 years, height: 164.2±10.5 cm, weight: 59.4±13.65 kg) and eight normal controls (age: 20±1.1 years, height: 164.8±6.1 cm, weight: 63.3±14.3 kg). Twenty-eight markers were used to track the motion of the lower limb. Each subject walked at self-selected pace and step over obstacles of three different heights (10, 20 and 30% of leg length) with each limb. Kinematic and kinetic data were measured with a 7-camera motion analysis system (Vicon, Oxford Metrics, U.K.) and two force plates (AMTI, U.S.A.). The clearance distances of big toe, heel and lateral malleolus markers were calculated. A model of the lower limb was used to calculate the angles and moments at the joints. In this paper, we report clearance distances and knee moments. Clearance distances and knee moments between limbs were compared using repeated t-test while independent t-test was used for between-group comparison. The significance level was set at 0.05.

RESULTS AND DISCUSSIONS

Compared to normal subjects, leading leg clearances of both affected and sound limbs in ACLD subjects were significantly smaller while in ACLR subjects those of the sound limbs were not significantly different from normal but those of the affected limbs were smaller, Fig.1. Leading leg clearances in affected limbs were significantly smaller than those in sound limbs for 20% and 30% conditions in ACLR group, and for 30% condition in ACLD group. This suggests that ACLD subjects are at higher risk in tripping over obstacles while ACL reconstruction could help to restore the subject's ability in successfully crossing over obstacles of lower heights. In ACLR subjects, the restored stability of the knee in the affected limb results in the close-to-normal clearance of the sound leading leg but the smaller obstacle clearance of the affected limb may be due to the partial restoration of the propriocetion of the joint.

Trailing leg clearances were indifferent among the groups and were found to be much larger than leading leg clearances. The high clearance for the trailing limb is needed to ensure safe obstacle-crossing when visual cue is not available and the results suggest that this mechanism is less sensitive to the structural and proprioceptive impairment of the ACL.

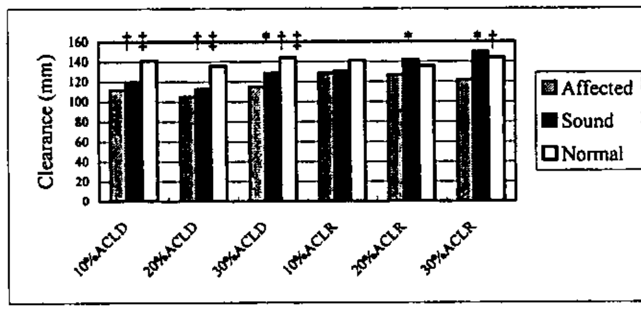

Fig. 1: Comparison of clearances in different conditions.
* Significant difference between affected and sound limbs
† Significant difference between affected and normal limbs
‡ Significant difference between sound and normal limbs

Knee extensor moments in the affected stance limbs were significantly bigger than those of the sound limbs when crossing obstacles of all heights in both ACL groups (ACLR: 14.8, 5.2 Nm, $p<0.004$; ACLD: 14.1, 5.7 Nm, $p<0.029$). This may be a result of muscular adaptation that tries to reduce anterior tibial translation in ACLD patients and protect the reconstructed ACL from excessive stress in ACLR patients.

Injury of the ACL affected joint proprioception and stability, resulting in insufficient information for the appropriate positioning of the leading leg and altered muscular control and force-bearing in passive structures in providing necessary stability of the body. The results of the present study suggest that ACLD patients are susceptible to tripping or falling in stepping over obstacles during locomotion and reconstruction of the ACL could reduce the risk.

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附錄

國家衛生研究院九十一年度整合性醫藥衛生科技研究計畫

前十字韌帶損傷及重建後膝關節之生物力學與神經肌肉適應之分析：

第一子計畫 — Biomechanical Modeling and Analysis of the Knee in Normal, ACL-Deficient and ACL-Reconstructed Patients

執行機構：台灣大學醫學工程學研究所

子計畫主持人：呂東武

研究人員：洪啓峰

Abstract

A subject-specific computer graphics-based modeling technique for the human knee joint was developed. The bones were modeled as rigid bodies, the ligaments as bundles of nonlinear tension bands and muscles as Hill-type force elements. The geometry of the force-bearing structures was described using data derived from CT and MR images. Knee laxity and maximum isometric muscle contraction tests on a young healthy subject were performed to provide data for model customization and validation. The model mechanical properties were customized to the subject by simulating drawer test that matched simulation results with experimental measurements. The model was also used to simulate passive knee flexion as well as the effects of hamstring action on the anterior tibial translation under a constant quadriceps force, both in good agreement with the literature. The model will be useful for a better understanding of the normal functions of the knee joint and its surrounding structures and hence provides necessary knowledge for ligament reconstruction in diseased or injured knees and the subsequent planning and evaluation of rehabilitation programs.

Keyword: knee biomechanics, ACL, computer model

Introduction

This component project is aimed to develop a three-dimensional computer model of the human knee joint, to be incorporated into an existing locomotor system model, for the study of the complicated mechanical interactions between the anterior cruciate ligament (ACL) and other force-bearing structures of the joint in normal, ACL-deficient and ACL-reconstructed patients during rehabilitation exercises and functional activities. It is hoped that the study will help enhance the current knowledge of the role of the ACL in controlling the mobility and stability of the knee during passive and dynamic multi-joint movements. This knowledge as well as the model will be helpful for future surgical treatment and rehabilitation of ACL injuries.

The mobility and stability of the knee joint are controlled by a complex interaction between the articular surfaces and the surrounding connective tissues including ligaments and muscles. During motion, the articular surfaces of the knee roll and slide upon each other while the ligaments and muscle tendons rotate about their origins and insertions on the bones. The positions and directions of the lines of actions of the forces transmitted by the structures and their lever arm lengths relative to the axis of rotation of the joint therefore vary systematically over the range of motion (Lu and O'Connor, 1996; 1997). Study of these interactions helps to establish a better understanding of the normal functions of the knee joint and its surrounding structures and hence provides necessary knowledge for ligament reconstruction in diseased or injured knees and the subsequent planning and evaluation of rehabilitation programs.

At present, knowledge of the interactions between the force-bearing structures at the knee comes mainly from *in vitro* studies (e.g. Markolf et al., 1990), limited number of *in vivo* measurements (e.g. Beynnon et al., 1995) and theoretical calculations (Kaufman et al., 1991; Morrison, 1970; Toutoungi et al., 2000). *In vitro* studies have provided some insights into the kinematic geometry and mechanics of the knee joint structures during passive movement (Herzog and Read, 1993; Rudy et al., 1996; Wilson et al., 1998) but motion and loading of the knee during voluntary dynamic movement is difficult to simulate in an experimental setup. Direct measurement of the *in vivo* kinematic and kinetic quantities is possible only through the use of invasive techniques such as bone pins for skeletal motion (e.g. Lafortune et al., 1992), implantable force transducers for tendon force measurement (e.g. Komi et al., 1987) and instrumented prostheses for the forces in bones (e.g. Lu et al., 1997). Therefore, due to ethical considerations and technical limitations, combined theoretical and experimental approaches have been a frequent choice for the determination of the kinematics and forces of the muscles, ligaments and bones of the knee joint during activities. This usually involves musculoskeletal models and noninvasive measurements. Motion analysis (or gait analysis) has been an efficient noninvasive method to acquire *in vivo* motion data of the human body segments and its joints, contributing to the understanding of human motions and the etiology of relevant diseases (Sutherland et al., 1972; Kadaba et al., 1990; Gage et al., 1995). However, the relative movement of the skin-markers with the underlying bone can cause errors in the estimated kinematics of the body segments and joints unless bone-pins are used (Lafortune et al., 1992; Lu et al., 1999).

Generally, analysis of the knee joint during functional movement is usually performed with two phases. Firstly, the lower limb is modeled as a multi-link system with simple mechanical joints such as a hinge knee, sometimes called "external model". The kinematics of the model is calculated with marker data from the gait laboratory and then used to determine the resultant forces and moments of each joint with kinetic data such as ground reaction forces measured from forceplates. The above-mentioned skin-bone relative movement is a major source of error in this phase. The second phase is to apply the calculated joint moments and forces onto another more detailed model of the knee joint, sometimes called "internal model", to calculate the forces in the force-bearing structures. The problem with this two-phase approach is that even though the internal knee model is very accurate, the mismatch between the detailed internal knee model and the oversimplified external one in the lower limb model can cause inaccuracies in the joint positions and the magnitudes and lines of action of the resultant forces and moments, leading to errors in the geometry and forces in the muscles, ligaments and articular contact surfaces. Recently, the responsible investigator suggested that detailed anatomical joint constraints should be included in the first phase of analysis not only to reduce effects of skin movement artifacts but also to eliminate the problem of knee model mismatch (Lu, 1999; Lu and O'Connor, 1998, 1999).

Several mathematical models of the knee have been proposed in the literature (Blankevoort and Huiskes, 1991; Blankevoort and Huiskes, 1996; Essinger et al., 1989; Li et al., 1999; Wismans et al., 1980; Shelburne and Pandy, 1997). Most of them calculated the motion of the knee and forces of the joint structures simultaneously using numerical methods including finite element analysis. Compared to the two-phase approach in motion analysis, where kinematic and mechanical analyses are performed separately and sequentially (Kinematics has to be found before mechanical analysis can be performed), these models are not suitable for the incorporation into gait models.

It is also noted that studies on the knee joint in the literature have been based on isolated knee joint models or below knee models (Andriacchi et al., 1996; Essinger et al., 1989; Kaufman et al., 1991; Morrison, 1970). The influence of the hip joint position on the lines of action of the bi-articular knee muscles has been largely ignored. Only equilibrium at the knee is considered so muscle forces calculated may not satisfy the equilibrium at the hip or the ankle during multi-joint movements. Therefore, for the study of knee mechanics during functional activities a complete model of the musculoskeletal locomotor system considering all the major joints in the limb should be used.

For theoretical calculations to be useful for clinical purposes, validation of the models and calculation procedures is necessary and has been a big challenge. Among the published mathematical models, some of the studies were validated by using in vitro data from isolated knee specimens (Blankevoort and Huiskes, 1996; Li et al., 1999) but few of them have been validated against living data directly measured (Lu et al., 1998). For the validation of the model-predicted mechanical behavior of the knee joint structures, in vitro experimental data are useful.

This component project is carried out in three years, each with different but closely related goals: (Year 1) development of a computer graphics-based model of the human knee joint; (Year 2) validation of the knee model with in vitro experiments; and (Year 3) incorporation of the knee model into an existing model of the human locomotor system to study of the mechanics of the ligaments during functional activities. During the past 10 months, the proposed research for Year 1 has been performed according to the plan and is described in this report.

Materials and Methods

A young healthy subject (24 yr; 176 cm; 74 Kg) participated in the study with informed consents. He was subjected to CT and MR scans while lying supine with knee extended (Fig. 1). The MR scan was performed with a surface coil and spanned from the medial the lateral extremes of the knee, enclosing a cubic viewing volume of approximately 192mm on each side. The MR images consisted of parallel digital images separated at 1.5 mm intervals with a resolution of 512x 416 pixels. The contours of the femur, tibia, patella, and ligaments were digitized within each image and reconstructed to obtain their three-dimensional geometry (Fig. 2).

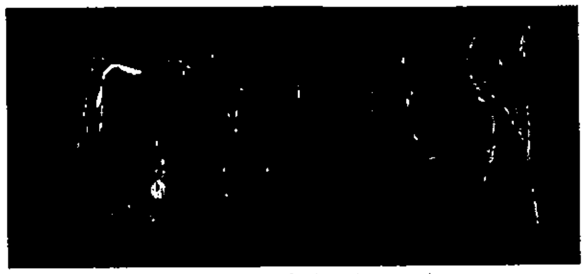

Fig. 1. MR images of the knee in transverse and sagittal planes.

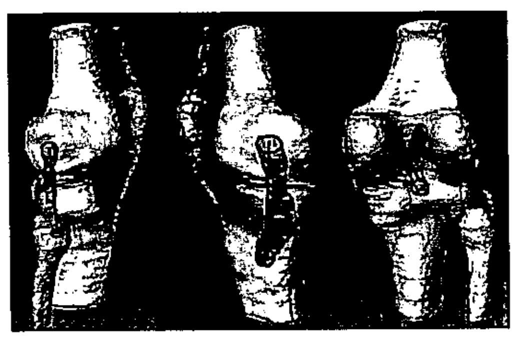

Fig. 2. The reconstructed geometry of the knee model and ligament attachments defined.

The laxities of the knees were measured using an arthrometer (KT2000, U.S.A.). The subject performed isometric quadriceps contraction with proper stabilization of the trunk and limbs in a gait laboratory while the tibia was constrained by a wire with one end around the ankle and the other fixed to the wall. A load cell in series with the wire was used to measure the force transmitted in the wire (Fig. 3). Passive infrared retroreflective markers were attached to the skin of the body segments for the description of their spatial positions. A video-based motion data acquisition system, Vicon 512

(Oxford Metrics Ltd., Oxford, England), was used to record the three-dimensional coordinates of the markers for subsequent analysis. The movements of the patients were also recorded using a digital camera to assist gait data interpretation (Fig. 4).

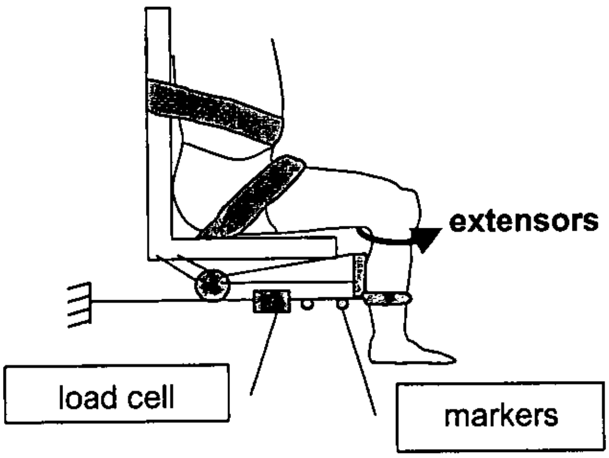

Fig. 3. Isometric quadriceps contraction testing device.

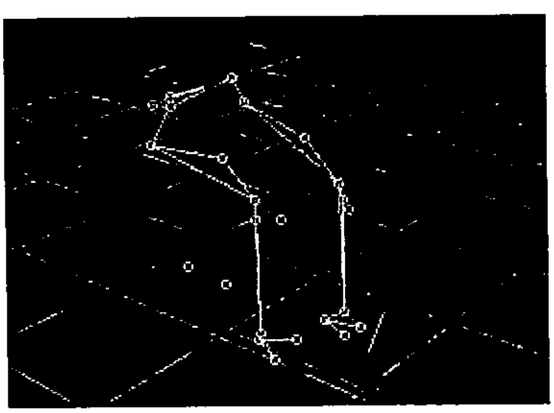

Fig. 4. Stick figure of the subject performing isometric quadriceps contraction at 90 deg of flexion. The movements of the segments were calculated with the measured marker positions (circles).

A subject-specific computer graphics-based model of the human knee, namely tibiofemoral and patellofemoral joints, was developed. Since the stiffness of the femur, tibia and patella is much greater than that of the relevant soft tissues, bones were assumed to be rigid in this model. The geometry of the bones and all other force-bearing structures were described using data derived from subject-specific CT and MR images. The ligaments of the knee, namely the ACL, PCL, MCL and LCL, were modeled as bundles of nonlinear tension bands, each connecting their origins on the femur and insertions on the tibia according to MRI-reconstructed models (Fig. 2). This modeling technique was capable of explaining the nonlinear relationship between the joint angles and knee laxity (Wismans et al., 1980; Blankevoort et al., 1991). Ligament strain is defined by $\varepsilon = (L - L_0)/L_0$, where L is the ligament length after deformation and L_0 the reference length of the ligament at which the ligament started to carry tensile force. The ligament force is quadratic function of the ligament strain when the ligament strain is less than a value of, where εl is a nonlinear strain level parameter and initially assumed to be 0.03. A linear force-displacement relationship is assumed when the ligament length is great than $2\varepsilon l$. Therefore, the force-displacement relationship of a ligament can be described using the following function (Blankevoort et al., 1991):

$$f = \begin{cases} \frac{1}{4} k \cdot \varepsilon^2 / \varepsilon l, & 0 \le \varepsilon \le 2\varepsilon l, \\ k(\varepsilon - \varepsilon l), & 2\varepsilon l \le \varepsilon, \\ 0, & \varepsilon \le 0, \end{cases} \qquad (1)$$

where k is a stiffness parameter. The muscles, including quadriceps and hamstrings, were modeled as Hill-type force elements.

Certain parameters of the model, namely the reference lengths of each ligament bundle element (l_i^r, where i = 1 to 6,representing the 6 ligament elements), the stiffness of ligament (k_j^l , where j = 1 to 4, representing the 4 ligaments) and nonlinear strain level parameter (εl), were modified through an optimization procedure since they are not easily accessible to experimental measurement. By doing so, the model could reflect accurately the mechanical properties of the force-bearing structures of the knee being studied. During the optimization process, the model parameters were taken as design variables and were systematically altered. For each set of model parameters, computer simulation of the knee at 20 degrees of flexion was performed to obtain the anterior-posterior tibial translation subjected to forces applied perpendicular to the tibia around the ankle obtained from the experiments. The optimization process stopped when the differences between the anterior-posterior tibial translations predicted by the model and those measured from the experiments were minimized. This objective function can be stated mathematically as follows

$$\min_{l_i^r, k_j^l, \varepsilon l} g = \left\{ \sum_{n=1}^{300} (T-t)^2 \right\}^{1/2}, \qquad (2)$$

where t and T are the model-calculated and measured anterior-posterior tibial translations, respectively.

After model customization, the model was used to simulate passive knee flexion and to study the effects of the hamstring action on the anterior tibial translation under a constant quadriceps force.

Results

The geometry of the subject-specific model was ensured by the CT and MR images of the subject while the mechanical properties of the model ligaments were achieved by simulating the knee laxity test (Figs. 5 and 6, Table 1). The optimum objective value was 2.395 for simulating the knee laxity test. The variations of most of the design variables were small, but the value of nonlinear strain level parameter increased from 0.03 to 0.0682878. The ligament forces in knee during anterior-posterior drawer test were shown in Fig. 9. The ligament forces of both anterior and posterior bundles of the ACL increased while subjected to anteriorly applied tibial load, and posterior bundles of the PCL increased with posteriorly applied tibial load (Fig 7). The subject-specific model was able to predict the coupled internal rotation and adduction in the knee during passive flexion observed by Wilson et al. (1998), Figs. 8 and 9. This suggests the knee has a single degree of mobility during passive knee flexion and extension. The model-calculated isometric quadriceps force at 90° knee flexion was found to be about 14 times the external tibial restraining force (Table 2). The model study of the effects of hamstring action on the anterior tibial translation found that the anterior tibial translation decreased with increasing hamstring force, and the anterior tibal displacement of knee of ACL-D patient would be greater that of normal (Fig. 10).

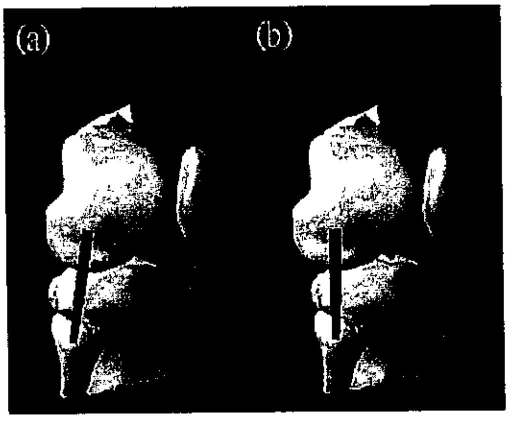

Fig. 5. Simulation of drawer test: tibia was (a) unloaded and (b) loaded by 159N anteriorly.

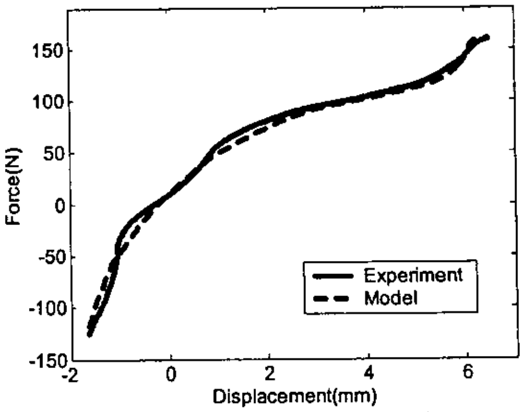

Fig. 6. Model-calculated and measured force-displacement curves during drawer test.

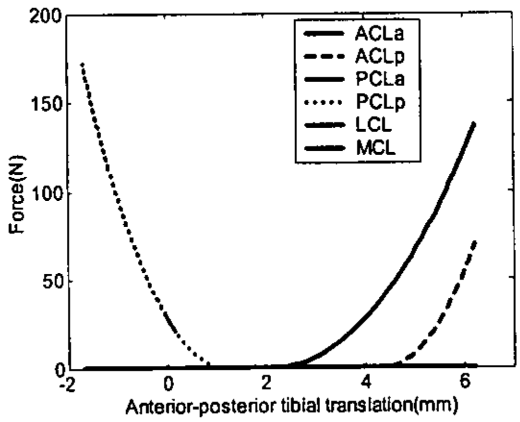

Fig. 7. The relationship of ligament forces verse anterior-posterior tibial translation during simulating the knee laxity test.

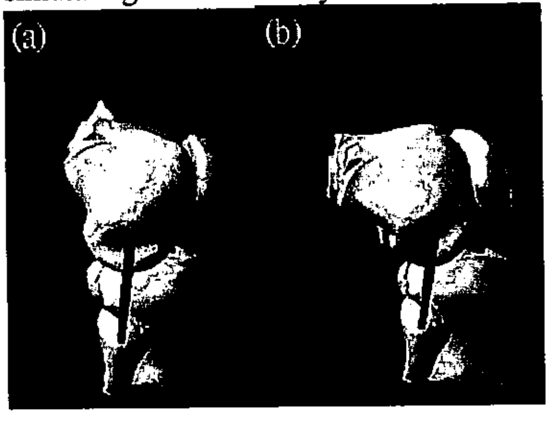

Fig. 8. Simulation of passive flexion of knee: the (a) 45° and (b)90° flexion at lateral side viewpoint.

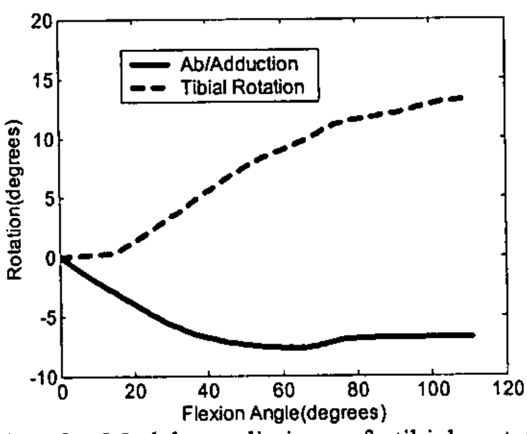

Fig. 9. Model prediction of tibial rotation relative to the femur as a function of flexion.

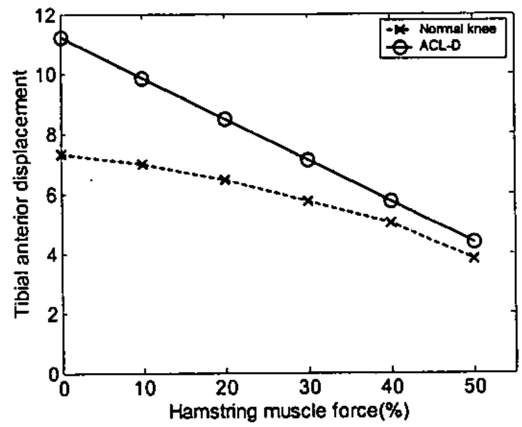

Fig. 10. The compensation of hamstring for anterior laxity.

Table 1 Knee ligaments and functional bundles with stiffness parameter, reference length and nonlinear strain level parameter.

Knee ligament	Ligament bundles	Stiffness parameter (N)	Optimum value (N)
ACL	Anterior bundle	5000	4999.89
	Posterior bundle	5000	4999.96
PCL	Anterior bundle	9000	9000.07
	Posterior bundle	9000	9000
LCL	Inferior bundle	6000	6000
MCL	Inferior bundle	8250	8250

Knee ligament	Ligament bundles	Reference length (N)	Optimum value (N)
ACL	Anterior bundle	34.9867	35.945
	Posterior bundle	21.9869	22.1017
PCL	Anterior bundle	29.278	29.29
	Posterior bundle	41.9149	41.9149
LCL	Inferior bundle	55.2759	55.2759
MCL	Inferior bundle	85.3603	85.3603

Nonlinear strain level parameter	Literature	Optimum value (N)
εl	0.03	0.0682878

Table 2 Ligament & contact forces of knee during isometric quadriceps contraction.

Ligament & Contact forces	Force(N)
External tibial restraining force	192.7
Model-calculated isometric quadriceps force	2713
Patellofemural joint	1449
Tibiofemural joint	2744
Patella tendon	1316
ACLa	37.382
PCLa	324.36
MCL	24.27

Discussion

A subject-specific knee modeling technique was developed and the model constructed with this technique for a young healthy subject was shown to be valid compared with data from experiments and the literature. The model will be helpful for further studies in the second and third part of this component project. It will be useful for a better understanding of the normal functions of the knee joint and its surrounding structures and hence provides necessary knowledge for ligament reconstruction in diseased or injured knees and the subsequent planning and evaluation of rehabilitation programs. The use of computer graphics-based approach provides a good opportunity for further application in surgical simulation, planning and computer-assisted surgery, for ACL injuries.

Acknowledgements

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國家衛生研究院九十一年度整合性醫藥衛生科技研究計畫

前十字韌帶損傷及重建後膝關節之生物力學與神經肌肉適應之分析

(Biomechanics and Neuromuscular Adaptation in ACL Patients)：

第二子計畫 － Gait Analysis Studies of ACL Injured and Reconstructed Knees

執行機構：中國醫藥學院物理治療學系

子計畫主持人：許弘昌

研究人員：林秀真、陳淑雅、趙偉鈞、高德昌、蔡宗遠

Introduction

This project is aimed to integrate the three component projects to investigate the short-term and long-term changes of gait variables and muscle activation pattern in ACL deficient (ACL-D) and reconstructed (ACL-R) patients during various functional activities. During the first year, we setup the integrated 3-dimensional motion analysis system including optic-tracking cameras, force platform system, and electromyographic (EMG) system. All the participants including ACL-R, ACL-D and healthy controls were examined in our motion analysis laboratory.

Both tibia and femur and their surrounding ligaments determine the passive motion of the knee joint. Muscles through the action of neurons thus control the dynamics of the joint, and work together with the passive components of the knee joint. The knee joint works sophisticatedly with the complex and interactive neuromusculoskeletal system [1]. Any injury of these elements will result in deterioration of knee joint. Anterior cruciate ligament (ACL) is one of the important elements that provide stability of knee joint, not only through the function of passive restrain of tibia anterior translation, but also through the sensory feedback of dynamic control of muscles [2-6]. Neuromuscular adaptation in the lower extremity has been found in anterior cruciate ligament (ACL) injured patients, with mechanical and electromyographic alterations such as reduced knee extensor moment and power, increased hamsrings EMG activity, and decreased muscle strength [7-15]. It is commonly believed that these alterations were demonstrated in various functional activities.

Nowadays, three-dimensional motion analysis has been extensively used in investigating the movement deviations during various functional activities in ACL patients. The quadriceps avoidance gait pattern proposed by Berchuck et al [7], which suggests a decreased external extension moment at the knee in stance, was found in the ACL-D patients and ACL-R patients [16]. Moreover, Wexler et al found greater flexion angles in midstance phase in ACL-D patient over time. Therefore, there was a special interest of studying in anterior translations and moment in sagittal plane for the ACL patients [20-22]. However, some phenomena were still considered controversial since there still were studies found different patterns in ACL-D [17, 18] and ACL-R patients [19]. Furthermore, the co-contraction of quadriceps and hamstring muscles could be an unsolved question in this analysis method as an indetermined problem [23]. Timoney et al reported that quadriceps activity performances in ACL-R patient improved after reconstruction surgery [19]. The alterations in the strength and duration of quadriceps and hamstring activities should be considered with the external moment changes.

The adaptations in performing functional activities in ACL-D and ACL-R patients were seldom been discussed, such as stepping over the obstacle, sit-to-stand, and stepping up and down stairs. Stepping over the obstacle, Chou et al reported that the flexion moment of hip joint of the trailing leg increased linearly with obstacle heights in young healthy subjects. Flexion moment of knee joint increased in the same fashion. Other parameters correspond to the height of obstacle are still controversial [24].

In describing sit-to-stand, five phases were defined by ground reaction force in the research of Kralj et al [25] and Millington et al [26]. It had been revealed that using a higher chair seat resulted in lower moment at knee and hip joints; lowering the chair seat increased the need for moment generation [27]. Joint moment is also influenced by speed of rising and the position of foot placement [28]. In this study, we consider about how the different heights of chair influence the activity of sit-to-standing in ACL-D and ACL-R patients.

Stair climbing is another common functional activity. It is a closed kinetic chain exercise and is thought to be an appropriate rehabilitation exercise for ACL-injured patients [29]. Andriacchi et al [30] reported that reduced knee flexion moments were not seen in ACL deficient patient, because less anterior shear force produced by quadriceps in larger knee flexion angles. Kowalk et al [16] investigate the kinetic changes during stair ascent in ACL deficient patients and after their reconstruction. They reported that the peak moment, power, and work in affected limb were not different with normal control in ACL deficient limb before reconstruction; however, they were significantly reduced after ACL reconstruction. They proposed these reductions were accommodated by increase in excursion, moment and power at the contralateral ankle joint.

In the previous literature, biomechanical analyses of above activities are focusing in healthy subjects, and there are few studies discussing the alterations in ACL-D or ACL-R patients. The recovery mechanism in ACL patients is still not clear with limited literature. Moreover, the forces of muscles and ligaments around knee joints were usually considered as a net force or net moment around knee joint. The use of dynamic electromyographic technique is one method to reveal the changes in muscular activation pattern in various function activities [31-33]. However, the substantial force-bearing and interactions between the muscle and ligament could be investigated mainly through *in vitro* studies [34, 37], computerized calculation with theoretical model [36, 37] and *in vivo* studies with invasive techniques [38-40]. With the advancements in the tools for studies of human locomotion, the most frequently used method involves placing markers on the skin of the segment being analyzed [41]. However, these kinds of the estimated kinematics of the body segments and joints with skin markers have seldom been validated with data directly measured.

Functional knee bracing has been a common method to enhance functional knee stability in these patients for the past three decades. The effects of bracing were examined in many aspects, such as kinematic and kinetic changes [42-45], sensory feedback [46, 47], strain behavior [48], and even in physiological parameter [49]. However, in recent study of Ramsey's *in vivo* 3-dimensional analysis, bracing the anterior cruciate results in only minor kinematic changes [40]. Contrast with the results of the dynamic EMG analysis studies, the bracing increases muscle activities in lower extremity [50-52]. In the past literature, the effects of wearing bracing were focused particularly in the restrain ability in the anterior-posterior translation between tibia and femur. Moreover, the long-term effects of bracing on ACL-R patients have not been reported in the literature. There is thus an urgent need in establishing complete knowledge on these effects.

The hypothesis of this study is that the changes the static stability of knee joint, which were followed by injury and reconstruction of ACL, will induced the compensation or adaptation in neuromuscular system in order to provide dynamic stability of knee joint. These changes can be disclosed by the thoroughly examination with three-dimensional motion analysis during various functional activities in the forms of joint kinematics, moment, ligament and muscle forces, muscle activation, etc. Therefore, the specific aims of this investigation in the first year are:

1. To investigate the differences of gait variables and muscle activation pattern in ACL deficient and reconstructed patients during functional activities.
2. To examine the effects of functional knee bracing in three anatomic planes with three-dimensional motion analysis.

Materials and Methods

Subjects

Ten patients with definite diagnosis of unilateral ACL deficiency and nine patients with bone-patella-bone autograft ACL reconstruction were recruited into this study from the Department of Orthopedics, China Medical College Hospital, Taichung. The participants had no low back problem or other knee pathology, age between 18 to 50 years old. People who had previously injured their lower extremities and had to confine to bed for more than a month, previous history of neurological disorder such as stroke, those who are (or anticipated to be) pregnant, were excluded. Those who have already shown significant arthritic changes were also excluded. Ten young healthy control subjects will also be recruited locally for comparison.

Instrumentation

Laxity of Knee joint. The anterior-posterior displacement was measured with the arthrometer (KT-2000, MEDmetric Co., U.S.A.). Many studies have shown the arthrometer to be valid, diagnostically accurate, and reliable method [53].

Functional knee braces. The DonJoy Gold point brace (Smith & Nephew DonJoy Inc.) was selected for studying the short-term and long-term effects for ACL patients. It was designed especially for cruciate ligament defect to protect knee joint and prevent excessive undesired movement between tibia and femur.

Three-dimensional motion analysis system. Seven infrared camera optic-tracking system (VICON, Oxford Metrics, U.K.) was used in this project to record the movement trajectory of each segment of lower extremity. The ground reaction forces (GRF) were measured with the force platform system (AMTI, Mass., U.S.A.) during level walking. One of them was used as second step of a three-step stair; each step was 18cm height, to collect the ground reaction forces of consecutive steps during stair activities.

Muscle activation pattern. The dynamic electromyographic signals were recorded (MA300, Motion Lab., U.S.A.) with surface electrode from gluteal maximus, rectus femoris, vestus medialis, vestus lateralis, biceps femoris, semitendinosus & semimembranosus, tibialis anterior, and medial gastrocnemius. Since the EMG data was used mainly to provide information of the gross activity of the muscles instead of localized muscle fibres, pre-amplified surface electrodes with good signal-to-noise ratio (SNR) will be used. Careful preparation of the skin and proper choice of the locations of the electrodes will be used to reduce cross-talks between muscles. The raw data were processed to linear envelop with computer programs (MATLAB software) to extract the pattern characteristics of muscle activation.

Procedure

Before the experiment, all participants were given a detailed description of this study and explanations of all their queries to their satisfactory prior to the experiment. Written consent was obtained from all volunteers. Investigator then recorded subjects' basic data, injury history and Lysholm scale questionnaire. The subject dresses in shorts, the anterior-posterior displacement was measured with arthrometer (KT-2000, MEDmetric Co., U.S.A.) from both knees, and each has three measurements. Eight surface electrodes of dynamic EMG were then fixed on the muscles of right leg of normal subject or affected limb of ACL subject. A marker system was developed to enable the measurement of the bony landmarks around the knee joint without being interfered by the use of braces. Before markers were attached, investigators test the maximum volunteer isometric contraction (MVIC) of hip extension, knee flexion, knee extension, ankle dorsiflexion, and ankle plantarflexion. These MVIC data are used to normalize the signal from motion trials.

Seven infrared camera optic-tracking system was set appropriately and calibrated before the experiment. During motion capture, subjects performed self-selected pace level walking at least 6 trials, stepping over the obstacle (at least 3 trials for each limb) with three obstacle heights of 10%, 20%, 30% of their leg length, sit-to-stand from three heights of chair (knee height, knee height +10cm, knee height -10cm), stair ascent (3 trials of each leg), and stair decent (3 trials of each leg). Each subject was fitted with suitable DonJoy Goldpoint braces on the right leg for normal subject and injured limb for ACL subject and performed level walking and stair trials. A model of the lower limb was used to calculate the forces and moments at the joints of the lower limb (Lu, 2001).

Three-dimensional gait variables were analyzed as dependent variables, including maximum joint angles, and maximum joint moments during activities. Reduced data were analyzed with paired *t*-test between brace conditions (wearing versus without wearing brace) and limbs (sound versus affected limb), and with ANOVA to reveal the differences between knee conditions (ACL-R, ACL-D, and normal). All statistical measures were tested with a significant level of 0.05.

Results and Discussion

Level walking

No significant alteration was found in normal subjects with brace on during level walking (Figure 1, 2). There is neither significant effect of bracing on joint angles and moments during level walking for ACL subjects, except bracing slight increased knee flexion angle range (56.0° to 61.7°) in injured limb of ACL-D subjects, and more external rotation of the lower leg (14.21° to 17.1°) in injured limb of ACL-R subjects. These results indicated that the immediate effects of bracing were not observable, however, it is not clear if there will be some long-term effects after using the brace for a period of time, that allows neuromuscular adaptation occurs.

Compared with normal, little kinematic deviation was found in ACL subjects. The alterations of joint angles occurred in increased hip internal rotation of injured limb in ACL-D group, decreased knee adduction of noninvolved limb, increased hip external rotation of injured limb and increased ankle plantarflexion of both limb in ACLR group. However, differences between injured and noninvolved limb were more significant in both ACL groups (Table 1), which indicate the asymmetry between injured and noninvolved limbs.

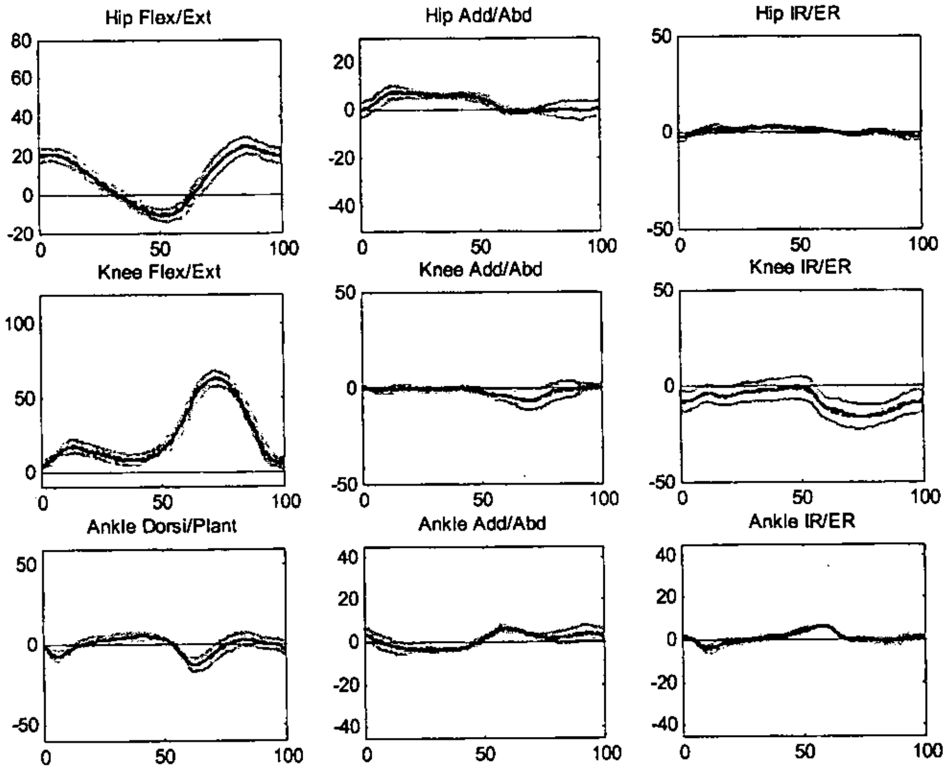

Figure 1. Average joint angles without (blue solid line) and with knee brace (red dash line) in normal subjects during level walking.

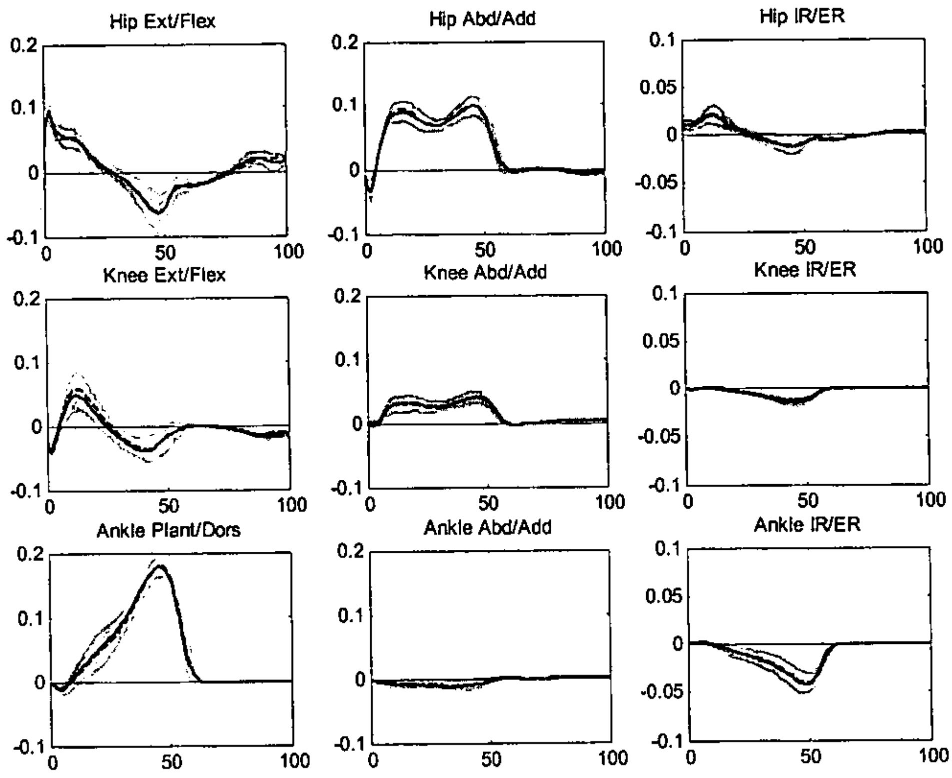

Figure 2. Average joint moments without (blue solid line) and with knee brace (red dash line) in normal subjects during level walking.

Table 1. Maximum joint angles during level walking.

	Normal		ACLD				ACLR			
			Noninvolved liml		Injured limb		Noninvolved liml		Injured limb	
	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)
Hip joint										
flexion	24.93	(4.53)	29.80	(4.20)	28.01	(4.24)	26.62	(7.02)	23.84	(5.89) *
extension	11.12	(3.32)	8.48	(3.10)	6.67	(3.74) *	8.22	(3.79)	7.89	(3.26)
Int. rotation	3.85	(1.73)	4.94	(1.89)	7.20	(4.65) a	4.29	(2.42)	5.17	(1.76)
Ext. rotation	3.25	(1.93)	5.36	(3.11)	5.82	(4.20)	5.08	(2.77)	8.64	(2.87) a
adduction	8.55	(1.43)	9.29	(2.67)	7.93	(2.78)	8.58	(2.54)	7.42	(2.88)
abduction	3.32	(2.21)	2.87	(2.29)	3.59	(3.81)	1.51	(1.54)	4.11	(2.96) *
Knee joint										
flexion	63.56	(5.38)	63.29	(2.20)	59.49	(9.43)	57.97	(12.93)	59.46	(2.96)
extension	-2.96	(4.24)	2.43	(4.10)	-0.62	(2.68) *	-2.19	(3.15)	-1.21	(4.22)
Int. rotation	0.10	(5.35)	4.09	(5.07)	0.11	(4.84) *	1.62	(3.29)	-0.54	(2.41)
Ext. rotation	17.65	(6.29)	13.01	(3.20)	13.96	(2.58)	15.63	(6.17)	15.67	(4.10)
adduction	2.15	(3.01)	1.79	(0.97)	2.67	(3.99)	-0.25	(2.01) b	1.09	(1.79) *
abduction	7.79	(4.63)	11.25	(5.17)	6.73	(6.99) *	10.02	(6.53)	11.08	(3.96)
Ankle joint										
dorsiflexion	5.92	(1.58)	5.74	(1.99)	6.79	(3.56)	5.61	(2.10)	5.73	(2.48)
plantarflexio	14.14	(4.47)	15.22	(3.26)	14.84	(3.42)	18.86	(2.74) b	18.57	(3.61) a
adduction	6.81	(3.24)	7.39	(3.10)	5.05	(3.12) *	7.47	(3.62)	6.31	(3.56)
abduction	4.49	(2.94)	6.11	(1.71)	4.33	(2.06)	5.30	(2.67)	4.79	(2.20)
inversion	6.87	(3.02)	7.40	(3.85)	8.07	(2.86)	7.60	(2.94)	6.82	(2.64)
eversion	4.43	(2.64)	6.47	(2.09)	5.10	(2.03)	3.71	(2.17)	5.73	(1.95) *

* p-value<0.05 in comparing the injured with noninvolved limb in ACL groups with paired t-test.

[a] p-value<0.05 in comparing the injured limb in ACL groups with normal limb with ANOVA.

[b] p-value<0.05 in comparing the noninvolved limb in ACL groups with normal limb with ANOVA.

Alterations in joint moments has similar finding that deviations from normal occurred in decreased hip flexor moment of injured limb in ACL-D group, decreased hip adductor moments of injured limbs in both ACL group, and decreased plantarflexor moments in both limbs of ACL-R group. There were several differences between injured and noninvolved limb were more significant in both ACL groups (Table 2). Interestingly, significant decrease of knee extensor moments between limbs in ACL-R subjects correlated to the EMG profiles finding. The EMG profiles of the normal and ACL-R subjects had no significant change after wearing knee braces, which also suggested that there was little immediate effect of bracing in terms of muscle activity. Compared to normal, ACL-R subjects showed decreased muscle activity in quadriceps both with and without bracing (Figure 3), indicating the presence of quadriceps avoidance. However, the detail EMG profiles have not been completely analyzed now, thus further reasoning cannot be done in ACLD subjects. More detail analysis in muscle activation profile will be done in the following two months.

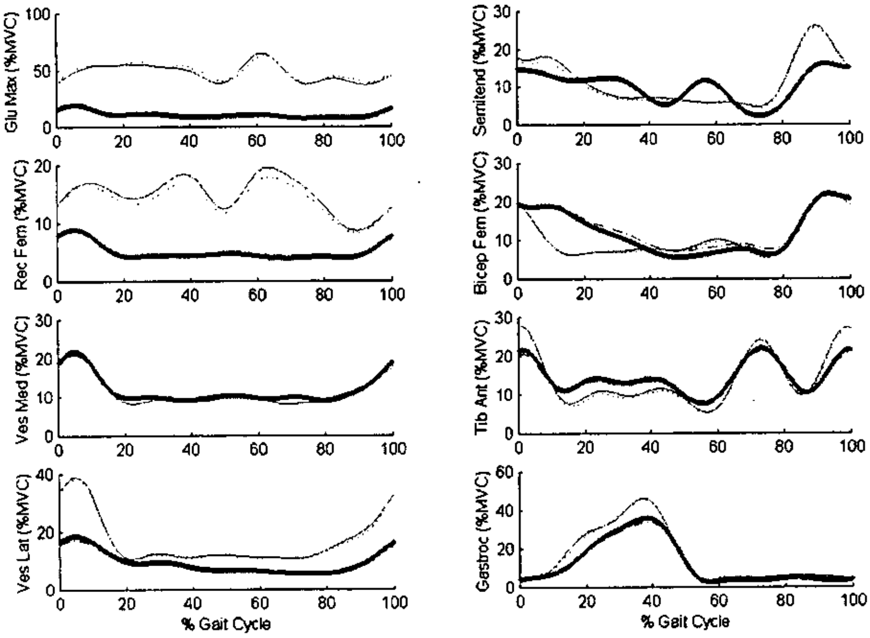

Figure 3. Comparisons of EMG linear envolop profiles of eight major leg muscles between normal (thin lines) and ACL-R (thick lines) subjects, with (dashdot lines) and without (solid lines) braces.

Table 2. Maximum joint moments during level walking (The values were normalized with leg length and body weight).

	Normal	ACLD		ACLR	
		Noninvolved liml	Injured limb	Noninvolved liml	Injured limb
	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)
Hip joint					
extensor	9.84% (1.51%)	9.64% (1.88%)	9.59% (1.87%)	9.21% (2.31%)	9.40% (2.52%)
flexor	6.86% (2.09%)	5.45% (2.10%)	4.44% (1.42%) a	6.74% (2.19%)	5.46% (1.63%) *
abductor	10.52% (1.33%)	10.22% (1.71%)	10.11% (0.89%)	9.69% (1.77%)	10.16% (1.90%)
adductor	3.73% (1.83%)	2.79% (2.37%)	1.34% (1.00%) a	2.47% (1.56%)	1.73% (1.25%) a
int. rotator	2.23% (0.96%)	1.79% (0.90%)	1.91% (0.80%)	2.28% (1.92%)	1.51% (0.68%)
ext. rotator	1.42% (0.72%)	1.45% (0.68%)	1.08% (0.74%)	1.92% (0.80%)	1.66% (1.72%)
Knee joint					
extensor	5.31% (2.70%)	3.44% (2.54%)	4.32% (1.85%)	5.98% (4.23%)	4.90% (4.92%) *
flexor	4.67% (1.57%)	4.90% (0.90%)	3.96% (0.91%) *	4.25% (0.91%)	4.16% (1.29%)
abductor	4.19% (1.00%)	3.50% (1.20%)	3.57% (1.33%)	4.20% (1.67%)	4.19% (1.16%)
adductor	0.61% (0.33%)	0.47% (0.56%)	0.42% (0.27%)	2.19% (5.54%)	0.29% (0.22%)
int. rotator	0.18% (0.16%)	0.12% (0.08%)	0.16% (0.17%)	0.66% (1.51%)	0.13% (0.17%)
ext. rotator	1.51% (0.44%)	1.24% (0.35%)	1.26% (0.49%)	1.82% (1.15%)	1.52% (0.60%)
Ankle joint					
plantarflexor	18.15% (1.84%)	16.62% (1.80%)	16.87% (0.62%)	16.75% (1.35%) b	16.54% (1.35%) a
dorsiflexor	1.18% (0.86%)	1.41% (0.38%)	1.13% (0.36%)	1.96% (1.97%)	0.66% (0.33%) *
evertor	0.23% (0.21%)	0.30% (0.66%)	0.29% (0.25%)	0.50% (0.39%)	0.23% (0.31%) *
invertor	1.26% (0.62%)	1.54% (0.51%)	1.59% (0.75%)	1.77% (2.61%)	1.25% (0.43%)
abductor	0.31% (0.16%)	0.34% (0.13%)	0.34% (0.14%)	0.33% (0.21%)	0.21% (0.09%)
adductor	4.22% (1.08%)	3.29% (0.66%)	3.56% (0.78%)	4.09% (1.01%)	3.71% (0.76%)

* p-value<0.05 in comparing the injured with noninvolved limb in ACL groups with paired t-test.

[a] p-value<0.05 in comparing the injured limb in ACL groups with normal limb with ANOVA.

[b] p-value<0.05 in comparing the noninvolved limb in ACL groups with normal limb with ANOVA.

Obstacle-crossing

The joint angles of the leading leg and joint moments of trailing limb during stance phase were analyzed. With increasing obstacle heights, ranges of motion of the three joints of the lower limb were quite different. The flexion angles of three joints of lower limb increased with obstacle heights, also maximum hip abduction and external rotation existed the same trend. However, joint moments of the stance limb during obstacle crossing were indifferent in all three conditions, except internal hip extensor moment slightly decreased with the obstacle heights increased. These results suggested that the strategy of crossing obstacles with different heights from the same position was mainly a regulation of joint angles of leading leg rather than regulation of joint moments of stance leg.

The maximum joint angles were not significant different in both ACL injured groups with normal limbs, except increased hip abduction and decreased knee flexion in ACL-D group in 10% condition (Table 3). The comparisons of joint moments in the stance leg have similar results, the only differences from normal occurred in increased maximum hip extensor moments of both limbs in ACL-D in 10% condition, and decreased maximum knee extensor moments of reconstructed knees in ACL-R group in all three conditions (Table 4, Figure 4). In 10% condition, the obstacle height was lowest and was not so challenging for the subjects. Therefore, it may reserved more capacity for variation of movement patterns. Conversely, the knee extensor moments of injured limbs in ACLR group were significant decreased in all three conditions, which suggested quadriceps avoidance phenomenon also existed in this activity. We can also observe the same decreased trend in knee extensor moments of the ACLD injured limbs (Figure 4). However, it did not reach the statistically significant level, which may result from the limited subject number.

Though few variables were found significant different with normal in both ACL groups, there existed many asymmetries between injured and noninvolved limbs (Table 3, 4). These asymmetries between limbs revealed accommodations occurred after the ACL injury. Since a safe and successful obstacle-crossing requires stability of the stance limb and sufficient foot clearance in the leading limb, which may not be guaranteed in ACL injured patients with impairment both the structural stability and sensory feedback of the joint. The cause of these asymmetries in ACL subjects could be either by the poor sensory feedback in the injured leading leg or by the degraded stability and sensory feedback of injured stance limb.

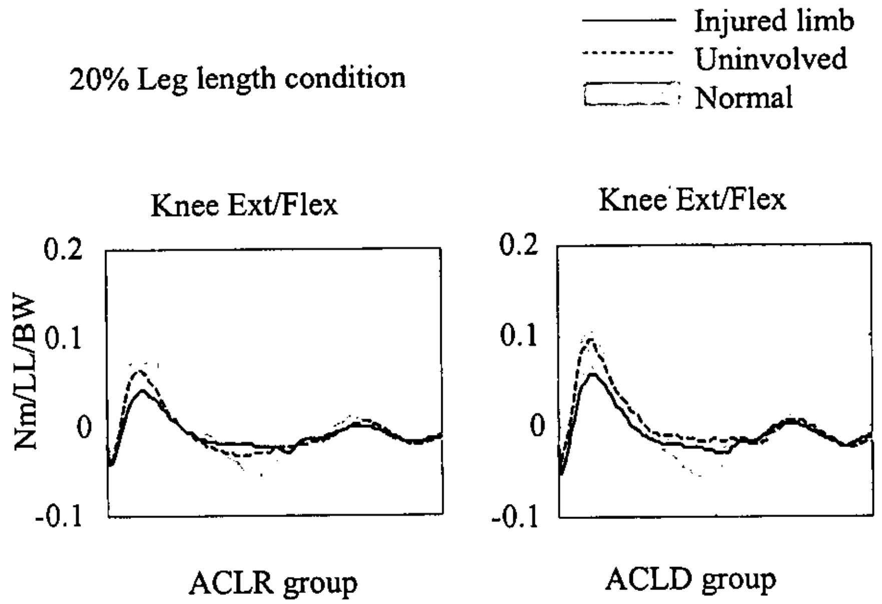

Figure 4. Comparison of knee extensor/flexor moments of ACLD subjects with normal subjects in 20% condition.

<u>Sit to stand</u>

The initial and terminal joint angle conditions were limited in experimental procedure, therefore, no significant different pattern was found between normal and ACL subjects. Compared with other functional activities, the moments on the coronal plane were obvious larger in values, such as value of maximum hip adductor moment was 0.1482 in this activity, 0.0154 during level walking, 0.0219 during obstacle-crossing, 0.0213 during stair ascent, and 0.0057 during stair decent. Analysis of the phases of this activity we found the point between acceleration and deceleration happened at 58.96% of the sit-to-stand cycle (Figure 3.5). It is comparable with the past literature (Kralj et al, 1990 and Millington et al, 1995).

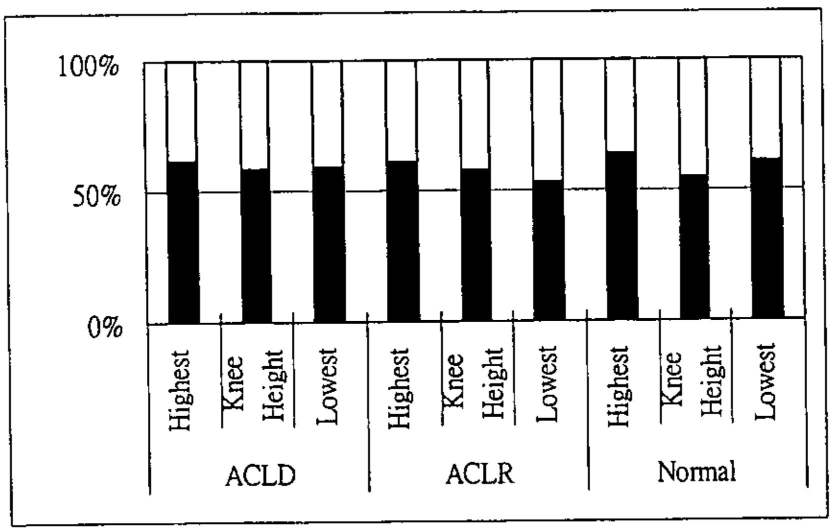

Figure 5. Phases of initiation (0%), acceleration, deceleration and end (100%) of sit-to-stand in three groups.

Table 3. Maximum joint angles during stepping over the obstacles with different heights.

		Normal		ACLD				ACLR			
				Noninvolved		Injured		Noninvolved		Injured	
	Heigh	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)
Hip joint											
	10%	50.18	(6.55)	46.53	(4.35)	47.63	(6.66)	50.67	(7.98)	52.65	(9.02)
flexion	20%	56.90	(8.48)	55.68	(6.84)	53.25	(19.27)	60.49	(6.12)	63.18	(10.05)
	30%	65.89	(8.43)	65.78	(6.27)	63.86	(19.43)	66.48	(5.60)	66.83	(10.14)
	10%	11.81	(4.43)	9.27	(2.46)	9.54	(1.86)	9.94	(1.99)	10.70	(2.40)
extension	20%	12.13	(5.18)	9.39	(2.78)	9.44	(2.35)	9.76	(2.09)	9.88	(3.05)
	30%	12.99	(4.96)	9.66	(2.40)	9.54	(2.70)	10.42	(3.26)	10.31	(3.29)
	10%	3.82	(1.20)	6.16	(5.73)	3.34	(3.92)	4.43	(2.03)	4.62	(1.79)
Int. rotation	20%	3.58	(1.48)	5.83	(5.82)	4.30	(5.57)	3.00	(2.53)	4.46	(3.63)
	30%	3.45	(1.62)	5.08	(5.02)	4.33	(6.21)	3.64	(2.86)	4.91	(3.41)
	10%	6.76	(2.36)	9.18	(3.68)	11.78	(3.11) a*	9.89	(3.06)	6.35	(3.92)
Ext.	20%	7.30	(3.15)	11.21	(4.90)	12.73	(6.08)	11.01	(3.66)	7.04	(4.96) *
	30%	9.24	(4.54)	14.21	(5.99)	14.10	(7.00)	12.06	(3.34)	8.37	(5.43) *
	10%	8.11	(2.19)	7.23	(2.73)	9.60	(5.63)	5.88	(2.47)	8.50	(2.65) *
adduction	20%	5.74	(1.71)	6.24	(3.07)	10.26	(11.94)	5.12	(1.70)	7.87	(4.17)
	30%	4.56	(1.21)	4.05	(2.70)	9.49	(14.02)	3.29	(2.10)	5.98	(4.96)
	10%	4.56	(2.68)	4.24	(3.04)	4.73	(3.08)	7.45	(3.54)	5.66	(3.12)
abduction	20%	6.09	(3.81)	4.05	(2.92)	5.69	(2.70)	8.03	(4.04)	5.76	(4.41)
	30%	9.12	(5.05)	5.95	(4.01)	8.10	(4.98)	9.44	(5.25)	6.93	(5.89) *
Knee joint											
	10%	94.04	(8.03)	77.49	(11.50) b	85.31	(5.48)	89.12	(7.28)	96.77	(6.14) *
flexion	20%	102.47	(10.00)	89.42	(14.78)	94.98	(11.12)	102.04	(7.62)	109.89	(4.64) *
	30%	113.42	(7.36)	100.26	(16.76)	111.57	(10.87)	108.03	(5.68)	112.64	(14.73)
	10%	-8.31	(4.52)	-8.83	(3.35)	-7.88	(2.39)	-8.52	(3.50)	-7.10	(3.50)
extension	20%	-7.74	(4.30)	-8.89	(3.57)	-6.90	(2.45)	-7.96	(4.48)	-7.20	(4.20)
	30%	-7.30	(4.74)	-9.67	(4.33)	-6.46	(4.29) *	-7.12	(3.58)	-6.19	(4.31)
	10%	1.53	(5.82)	1.52	(5.16)	4.26	(3.68)	1.17	(3.91)	3.92	(4.43)
Int. rotation	20%	1.97	(5.94)	3.02	(4.95)	4.13	(2.98)	-0.37	(3.58)	4.93	(3.69) *
	30%	2.32	(5.43)	3.49	(4.95)	5.69	(6.31)	1.54	(2.11)	4.25	(2.70) *
	10%	17.06	(5.75)	14.82	(2.62)	16.09	(5.71)	15.91	(3.91)	18.77	(6.76)
Ext.	20%	18.36	(4.08)	15.63	(3.67)	17.40	(6.01)	18.19	(4.67)	21.53	(9.24)
	30%	19.13	(4.14)	18.41	(7.44)	19.20	(6.20)	17.43	(5.23)	21.52	(7.92)
	10%	2.77	(4.17)	3.62	(7.98)	-0.81	(1.56)	0.38	(3.86)	2.60	(7.26)
adduction	20%	4.01	(4.94)	0.68	(5.81)	-0.90	(1.37)	1.48	(5.42)	3.14	(8.51)
	30%	7.48	(7.54)	5.45	(10.10)	1.37	(3.59)	3.03	(7.44)	5.13	(9.49)
	10%	7.97	(5.39)	9.77	(8.32)	11.68	(4.21)	11.87	(6.00)	10.93	(7.11)
abduction	20%	8.13	(4.95)	11.48	(6.40)	13.07	(3.90)	11.40	(5.18)	12.51	(8.15)
	30%	6.72	(5.30)	8.86	(6.90)	12.16	(4.90)	10.09	(5.62)	10.90	(7.84)
Ankle joint											
	10%	10.51	(2.89)	11.98	(2.56)	12.08	(2.31)	12.93	(5.04)	12.12	(4.60)
dorsiflexion	20%	10.53	(3.91)	14.37	(2.97)	12.38	(3.04)	14.09	(4.46)	12.32	(4.87)
	30%	11.10	(4.76)	16.02	(5.18)	15.83	(3.23)	15.24	(5.61)	12.37	(4.86)
	10%	14.12	(2.46)	18.39	(5.00)	16.35	(3.10)	19.15	(5.47)	19.49	(5.02)
plantarflexi	20%	16.86	(6.13)	17.55	(3.89)	17.76	(3.05)	20.71	(4.41)	21.88	(6.17)
	30%	18.02	(6.59)	19.47	(7.04)	22.65	(3.72)	21.38	(4.80)	21.12	(6.14)
	10%	6.67	(3.12)	6.86	(2.61)	9.95	(4.29)	7.98	(4.74)	8.38	(5.02)
adduction	20%	8.29	(2.71)	6.41	(2.95)	9.28	(2.41) *	7.95	(4.17)	8.01	(3.79)
	30%	8.35	(3.15)	6.82	(3.50)	10.92	(5.44) *	8.53	(4.95)	8.97	(4.00)
	10%	3.00	(2.67)	5.32	(2.31)	5.09	(3.08)	3.59	(2.51)	5.31	(4.81)
abduction	20%	3.37	(3.84)	4.76	(2.90)	3.27	(2.22) *	2.70	(3.14)	5.04	(4.19)
	30%	2.82	(3.74)	9.71	(15.59)	3.67	(3.67)	3.29	(3.05)	4.34	(4.74)
	10%	6.72	(3.46)	7.81	(3.20)	8.20	(3.74)	10.24	(3.39)	9.49	(2.88)
inversion	20%	7.54	(3.83)	7.46	(1.84)	8.06	(3.11)	9.13	(3.60)	8.67	(4.05)
	30%	8.07	(4.41)	6.58	(3.30)	9.32	(5.24) *	9.58	(3.25)	8.96	(4.33)
	10%	5.42	(2.78)	5.91	(3.26)	5.49	(1.63)	5.24	(3.11)	3.91	(3.36)
eversion	20%	4.52	(3.43)	5.47	(3.04)	4.54	(2.13)	4.52	(3.21)	3.73	(3.15)
	30%	4.26	(3.02)	5.57	(4.02)	4.58	(2.92)	3.75	(3.13)	3.39	(3.97)

* p-value<0.05 in comparing the injured with noninvolved limb in ACL groups with paired t-test.

a p-value<0.05 in comparing the injured limb in ACL groups with normal limb with ANOVA.

b p-value<0.05 in comparing the noninvolved limb in ACL groups with normal limb with ANOVA.

Table 4. Maximum joint angles during stepping over the obstacles with different heights.

		Normal	ACLD		ACLR	
			Noninvolved	Injured	Noninvolved	Injured
	Heigh	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)
Hip joint						
extensor	10%	8.06% (1.83%)	12.41% (3.62%) b	13.31% (3.56%) a	10.11% (3.20%)	9.87% (2.81%)
	20%	9.61% (2.18%)	10.75% (3.83%)	12.46% (3.84%)	8.96% (1.67%)	10.66% (3.31%)
	30%	9.03% (3.30%)	11.54% (3.86%)	13.00% (3.05%)	9.98% (2.37%)	10.89% (2.73%)
flexor	10%	4.68% (2.62%)	5.34% (2.92%)	3.32% (1.96%) *	5.22% (1.92%)	3.97% (1.62%)
	20%	3.92% (1.94%)	4.34% (3.15%)	2.85% (1.49%)	4.11% (1.27%)	3.16% (1.63%)
	30%	3.21% (1.51%)	3.35% (1.47%)	2.70% (1.07%)	3.43% (1.25%)	3.12% (1.33%)
abductor	10%	11.76% (3.15%)	12.77% (2.72%)	11.57% (1.89%)	11.13% (1.93%)	10.77% (1.65%)
	20%	12.68% (2.84%)	12.40% (3.81%)	11.90% (2.28%)	11.36% (1.95%)	11.48% (2.20%)
	30%	12.35% (2.78%)	13.01% (1.83%)	11.06% (1.71%) *	11.78% (1.67%)	11.98% (2.26%)
adductor	10%	3.16% (1.77%)	2.13% (0.86%)	1.90% (1.13%)	2.08% (1.77%)	1.53% (0.65%)
	20%	3.10% (2.23%)	2.01% (1.41%)	2.00% (1.88%)	2.09% (1.68%)	1.84% (1.16%)
	30%	2.57% (1.96%)	2.17% (1.74%)	1.64% (1.09%)	2.51% (2.06%)	1.95% (1.24%)
int.	10%	2.18% (0.96%)	2.42% (1.64%)	2.35% (0.73%)	2.04% (1.12%)	1.57% (0.67%)
	20%	2.92% (1.27%)	2.73% (1.45%)	2.54% (1.15%)	1.96% (1.01%)	1.86% (0.91%)
	30%	2.45% (1.19%)	2.74% (1.48%)	2.55% (1.03%)	2.21% (0.96%)	2.11% (1.00%)
ext.	10%	1.46% (0.74%)	1.42% (0.51%)	1.02% (0.64%)	1.39% (0.88%)	0.88% (0.43%) *
	20%	1.12% (0.73%)	1.53% (1.96%)	0.85% (0.35%)	1.26% (0.72%)	0.78% (0.41%) *
	30%	0.99% (0.55%)	1.50% (2.01%)	0.93% (0.79%)	1.13% (0.56%)	0.86% (0.60%)
Knee joint						
extensor	10%	8.57% (2.24%)	8.34% (4.67%)	5.76% (1.87%)	4.95% (2.27%)	3.07% (2.32%) a*
	20%	8.75% (2.24%)	9.94% (6.70%)	5.96% (1.78%)	6.34% (2.94%)	3.82% (2.66%) a
	30%	9.45% (1.28%)	10.99% (7.75%)	7.05% (4.52%)	6.41% (2.93%)	3.82% (1.82%) a*
flexor	10%	3.51% (1.01%)	5.01% (1.49%)	5.04% (1.55%)	4.45% (1.34%)	4.10% (1.25%)
	20%	4.68% (1.54%)	4.87% (1.36%)	5.37% (1.90%)	4.56% (0.53%)	4.61% (1.36%)
	30%	4.26% (1.45%)	5.34% (1.32%)	5.45% (2.06%)	4.62% (1.01%)	4.79% (1.13%)
abductor	10%	5.02% (1.40%)	4.80% (1.70%)	4.17% (1.55%)	5.15% (1.97%)	4.45% (1.86%)
	20%	5.10% (1.67%)	4.71% (2.40%)	3.97% (1.47%)	4.75% (1.54%)	4.58% (1.89%)
	30%	4.97% (1.96%)	4.95% (2.70%)	4.31% (1.88%)	4.68% (1.46%)	4.46% (2.17%)
adductor	10%	0.59% (0.35%)	0.79% (0.82%)	0.42% (0.31%)	0.73% (0.62%)	0.17% (0.30%) *
	20%	0.67% (0.30%)	1.94% (4.58%)	0.51% (0.39%)	0.50% (0.43%)	0.30% (0.30%)
	30%	0.66% (0.22%)	0.81% (0.39%)	0.58% (0.34%)	0.74% (0.54%)	0.50% (0.61%)
int. rotator	10%	0.28% (0.17%)	0.29% (0.16%)	0.23% (0.11%)	0.19% (0.24%)	0.07% (0.06%)
	20%	0.24% (0.20%)	0.28% (0.24%)	0.22% (0.14%)	0.20% (0.23%)	0.28% (0.30%)
	30%	0.29% (0.24%)	0.35% (0.39%)	0.25% (0.20%)	0.20% (0.18%)	0.48% (1.15%)
ext.	10%	1.66% (0.48%)	1.52% (0.48%)	1.38% (0.60%)	1.76% (0.85%)	1.39% (0.81%)
	20%	1.60% (0.74%)	1.99% (1.25%)	1.35% (0.72%)	1.53% (0.63%)	1.38% (0.91%)
	30%	1.54% (0.76%)	1.41% (0.72%)	1.33% (0.75%)	1.35% (0.64%)	1.28% (1.06%)
Ankle joint						
plantarflex	10%	19.71% (2.97%)	18.70% (3.18%)	18.51% (3.51%)	18.42% (2.21%)	16.87% (1.95%) *
	20%	20.33% (4.64%)	19.83% (4.83%)	20.10% (5.03%)	17.78% (2.23%)	16.89% (2.85%)
	30%	20.38% (4.68%)	18.94% (3.76%)	19.55% (3.08%)	17.56% (2.41%)	16.74% (2.96%)
dorsiflexor	10%	1.36% (1.22%)	1.59% (0.51%)	1.04% (0.70%) *	1.90% (1.70%)	0.93% (0.41%)
	20%	1.39% (0.98%)	2.82% (4.25%)	1.00% (0.53%)	1.31% (0.55%)	0.78% (0.47%) *
	30%	1.65% (1.24%)	1.27% (0.67%)	1.03% (0.74%)	1.24% (0.53%)	0.73% (0.43%) *
evertor	10%	0.40% (0.41%)	0.50% (0.79%)	0.41% (0.43%)	0.88% (0.81%)	0.20% (0.22%) *
	20%	0.30% (0.43%)	0.45% (0.71%)	0.60% (0.54%)	0.53% (0.52%)	0.56% (0.60%)
	30%	0.30% (0.32%)	0.49% (0.77%)	0.39% (0.47%)	0.53% (0.59%)	0.19% (0.26%)
invertor	10%	1.25% (0.69%)	1.78% (0.65%)	1.55% (0.66%)	1.18% (0.55%)	1.43% (0.50%)
	20%	1.64% (0.72%)	2.42% (1.73%)	2.02% (0.90%)	1.24% (0.70%)	1.73% (1.56%)
	30%	1.56% (0.57%)	1.60% (0.61%)	1.49% (0.65%)	1.39% (0.66%)	1.84% (1.39%)
abductor	10%	0.38% (0.32%)	0.45% (0.19%)	0.37% (0.17%)	0.32% (0.14%)	0.28% (0.13%)
	20%	0.43% (0.27%)	0.96% (1.56%)	0.40% (0.12%)	0.32% (0.12%)	0.38% (0.38%)
	30%	0.48% (0.34%)	0.46% (0.20%)	0.38% (0.21%)	0.34% (0.15%)	0.62% (1.09%)
adductor	10%	4.48% (1.52%)	4.31% (1.87%)	4.44% (1.11%)	4.49% (1.47%)	4.60% (1.52%)
	20%	4.87% (1.55%)	3.98% (1.69%)	3.71% (1.03%)	4.33% (1.27%)	4.44% (1.09%)
	30%	5.04% (1.72%)	4.27% (1.95%)	4.41% (1.45%)	4.79% (1.22%)	5.22% (2.13%)

* p-value<0.05 in comparing the injured with noninvolved limb in ACL groups with paired t-test.

a p-value<0.05 in comparing the injured limb in ACL groups with normal limb with ANOVA.

b p-value<0.05 in comparing the noninvolved limb in ACL groups with normal limb with ANOVA.

Stair Ascent

The kinematics and kinetics during stance phase of stair activities were analyzed. There was no statistically significant change in the joint kinematics and kinetics after wearing a brace in normal subjects, except slight decreased the maximum hip external rotation (7.28 to 6.42), knee flexion (62.43 to 61.44) and knee internal rotation (0.25 to -1.36). In fact, these changes might be trivial to sense and could resulted from restriction or unfamiliarity of the brace. Bracing mainly decreased ROM of hip and knee joints in sigittal plane in injured limbs of both ACL subjects, which decreased the maximum hip flexion angle decreased from 60.8 to 53.43 in ACLD, from 52.94 to 50.29 in ACLR group, and knee flexion angles decreased from 58.55 to 53.17 in ACLD and from 56.76 to 54.18 in ACLR group. However, no significant was change found in maximum joint moments in sagittal plane. The only changes occurred in decreased maximum hip external rotator moment of both ACL injured limbs.

Significant differences in ANOVA analysis between limb conditions were found in several kinematic and kinetic variables. Compared ACL injured limbs with normal, both ACL groups existed obvious decreased knee extensor moments (Figure 6). ACLD noninvolved limbs Compared with normal limbs also had alterations of increased hip flexion angle, hip external rotation angle, and increased ankle dorsiflexor moment (Table 5).

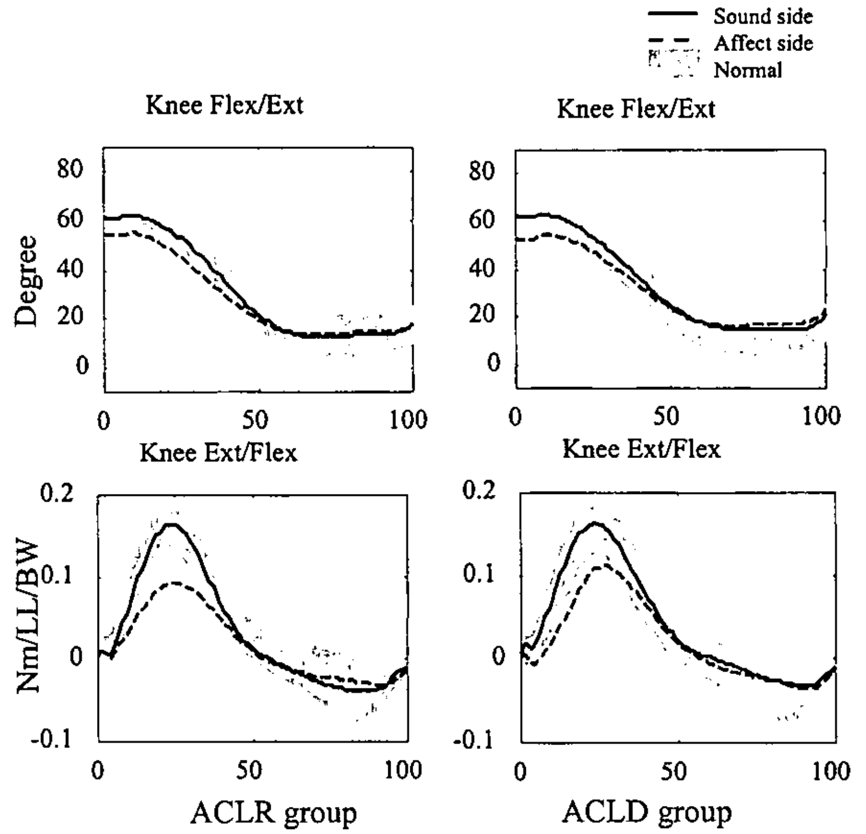

Figure 6. Comparison of knee flexion/extension angle and extensor/flexor moments of ACL subjects with normal subjects in stance phase of stair ascent.

The ACL injured limbs were significantly differed from noninvolved limbs. There were noticeable decreased maximum knee flexion angles and knee extensor moments in both ACLD groups (Figure 6). Besides, in the kinematic variables, smaller hip external rotation, knee internal rotation, and larger hip internal rotation in ACLD group; and smaller hip flexion, and knee abduction, larger hip abduction and ankle dorsiflexion in ACLR groups were found (Table 5). In the kinetic variables, smaller hip abductor moment in ACLD group, and larger hip extensor, smaller, knee flexor, internal rotator, ankle plantarflexor and ankle abductor moments in ACLR group were found (Table 6).

Table 5. Maximum joint angles in stance phase of stair ascent

Joint angles	Normal		ACLD				ACLR			
			Noninvolved liml		Injured limb		Noninvolved liml		Injured limb	
	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)
Hip joint										
flexion	52.74	(5.31)	60.85	(7.11) b	57.11	(7.25)	56.28	(4.63)	51.61	(4.46) *
extension	2.15	(5.12)	-2.98	(4.34)	-3.50	(5.87)	0.08	(4.27)	0.97	(3.47)
Int. rotation	2.73	(2.55)	0.38	(4.47)	3.08	(4.50) *	1.56	(4.01)	3.41	(2.49)
Ext. rotation	6.85	(2.72)	12.37	(5.50) b	7.96	(5.98) *	6.90	(4.15)	6.22	(2.24)
adduction	6.87	(4.61)	9.36	(7.51)	7.38	(6.84)	7.99	(4.94)	7.20	(3.19)
abduction	6.88	(4.13)	5.73	(1.88)	5.92	(4.04)	4.99	(3.66)	7.42	(4.02) *
Knee joint										
flexion	61.94	(6.28)	64.57	(6.30)	55.86	(11.05) *	63.27	(5.16)	55.47	(7.83) *
extension	-10.34	(8.27)	-12.00	(5.22)	-13.91	(5.08)	-10.61	(3.76)	-11.01	(3.66)
Int. rotation	-0.55	(8.15)	4.63	(7.46)	0.03	(5.84) *	1.42	(4.50)	-2.77	(5.74)
Ext. rotation	13.66	(6.53)	10.75	(7.23)	10.71	(5.00)	14.73	(3.22)	13.83	(5.73)
adduction	4.06	(5.30)	0.27	(3.88)	3.01	(9.39)	0.63	(7.58)	1.73	(4.88)
abduction	5.37	(4.15)	10.60	(7.35)	8.38	(7.87)	9.79	(5.70)	6.00	(3.95) *
Ankle joint										
dorsiflexion	13.75	(6.60)	15.09	(4.53)	12.80	(6.21)	14.88	(2.53)	10.47	(4.10) *
plantarflexion	15.15	(6.52)	17.01	(5.38)	16.00	(4.80)	18.22	(5.77)	16.29	(2.03)
adduction	6.16	(2.71)	8.42	(5.94)	6.23	(4.45)	6.49	(3.32)	6.81	(2.68)
abduction	4.65	(2.78)	5.15	(5.62)	4.71	(4.77)	2.80	(2.70)	4.04	(1.79)
inversion	7.23	(4.90)	10.42	(4.84)	8.18	(3.94)	6.73	(4.47)	6.96	(3.55)
eversion	5.89	(5.05)	8.19	(3.71)	6.69	(5.51)	4.60	(3.82)	5.83	(2.93)

* p-value<0.05 in comparing the injured with noninvolved limb in ACL groups with paired t-test.
[a] p-value<0.05 in comparing the injured limb in ACL groups with normal limb with ANOVA.
[b] p-value<0.05 in comparing the noninvolved limb in ACL groups with normal limb with ANOVA.

Table 6. Maximum joint moments in stance phase of stair ascent

	Normal	ACLD		ACLR	
		Noninvolved liml	Injured limb	Noninvolved liml	Injured limb
	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)
Hip joint					
extensor	10.32% (2.57%)	9.38% (2.95%)	10.77% (2.37%)	9.50% (2.10%)	11.44% (2.71%) *
flexor	2.10% (1.42%)	2.47% (2.53%)	1.78% (1.85%)	2.74% (2.16%)	1.39% (0.84%) *
abductor	9.58% (2.27%)	9.94% (1.99%)	8.49% (0.90%) *	9.09% (2.31%)	9.48% (1.98%)
adductor	3.17% (1.50%)	1.83% (1.17%)	2.18% (1.35%)	2.76% (2.37%)	1.95% (1.33%)
int. rotator	4.26% (1.13%)	4.68% (1.30%)	4.11% (1.14%)	3.91% (0.56%)	3.58% (0.92%)
ext. rotator	0.90% (0.63%)	0.93% (0.60%)	0.57% (0.59%)	0.86% (0.40%)	0.69% (0.50%)
Knee joint					
extensor	16.89% (2.34%)	18.43% (2.45%)	11.91% (3.96%) a*	17.57% (2.20%)	10.08% (3.54%) a*
flexor	5.22% (2.94%)	4.10% (0.99%)	4.56% (2.33%)	5.08% (1.33%)	4.06% (1.00%) *
abductor	4.18% (1.30%)	4.08% (2.50%)	3.11% (1.73%)	3.61% (1.89%)	3.50% (1.21%)
adductor	2.35% (1.18%)	2.18% (2.38%)	2.57% (2.41%)	2.41% (1.17%)	1.60% (1.28%)
int. rotator	0.77% (0.39%)	0.58% (0.30%)	0.89% (0.83%)	0.63% (0.18%)	0.39% (0.28%) *
ext. rotator	1.28% (0.51%)	1.22% (0.54%)	1.00% (0.69%)	1.25% (0.87%)	1.10% (0.45%)
Ankle joint					
plantarflexor	19.72% (1.62%)	19.37% (2.23%)	19.05% (1.80%)	19.30% (2.43%)	17.99% (1.98%) *
dorsiflexor	0.15% (0.22%)	0.63% (0.53%) b	0.42% (0.43%)	0.26% (0.50%)	0.17% (0.33%)
evertor	0.76% (0.50%)	0.83% (0.94%)	0.85% (0.83%)	0.92% (0.84%)	0.54% (0.40%)
invertor	1.51% (0.72%)	1.88% (1.14%)	1.92% (1.22%)	1.76% (0.75%)	1.59% (0.63%)
abductor	0.15% (0.09%)	0.39% (0.51%)	0.37% (0.37%)	0.25% (0.19%)	0.11% (0.10%) *
adductor	4.01% (1.43%)	3.68% (1.65%)	3.05% (1.46%)	4.38% (1.38%)	4.03% (0.95%)

* p-value<0.05 in comparing the injured with noninvolved limb in ACL groups with paired t-test.
[a] p-value<0.05 in comparing the injured limb in ACL groups with normal limb with ANOVA.
[b] p-value<0.05 in comparing the noninvolved limb in ACL groups with normal limb with ANOVA.

Stair Decent

There was no bracing effect found in normal and ACLR subjects both in kinematics and kinetics during stair decent. Alterations after wearing brace were trivial in values, and occurred at decreased hip adduction (6.45 to 5.26) and increased ankle inversion (8.89 to 12.27) angles.

Compared to normal kinematics, ACLR group demonstrated significant decreased maximum knee flexion angle in injured limbs, while ACLD subjects exist increased hip flexion and ankle plantarflexion angles in the noninvolved limbs (Table 6). The changes of joint moments mainly occurred in sagittal plane that notable decreased of knee extensor and ankle plantarflexor moments in injured limbs of both ACL groups. In coronal plane, hip abductor moment was significantly decreased in injured limbs of both ACL groups, and decreased knee abductor and internal rotator moments were found in injured limbs of ACLD subjects as well. On the other hand, noninvolved limbs of both ACL groups demonstrate hip flexor moment, and an increased knee extensor in ACLD group (Table 7).

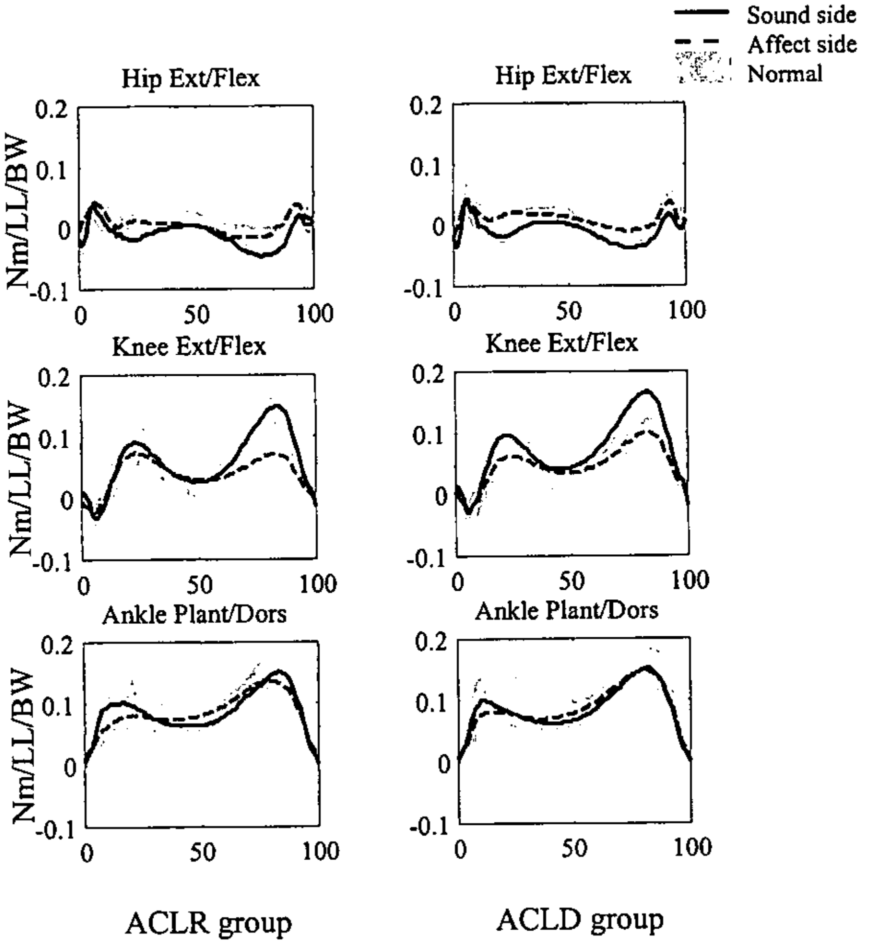

Figure 7. Comparison of moments in sagittal plane of ACL subjects with normal subjects in stance phase of stair decent.

The kinematic changes between injured and noninvolved limbs in ACL groups occurred mainly at knee and ankle joint movements in sagittal plane. Injured limbs had smaller knee flexion and ankle plantarflexion angles. Correspondingly, significant decreased ankle plantarflexor, and knee extensor were found in ACL injured limbs. In addition, smaller hip adduction angles, larger knee adduction angles, and smaller hip abductor moments were found in injured limbs of both ACL groups (Table 7, 8). Increased flexor moments in noninvolved limbs of both ACL groups might indicate a compensatory mechanism that ACL injured subjects might shift their body weights earlier and more forward to the noninvolved limbs during stair decent.

The strain of ACL during stair climbing [29] was not significant larger than other exercises. Sagittal plane knee translation measured with electrogoniometry during stair-climbing [54] even showed smaller displacement and higher muscle activity in ACL-deficient patients during closed kinetic chain exercises than open kinetic chain exercise. These results demonstrated that stair activities were a safe exercise for ACL patients. However, while considering the decreases in joint moments and ranges of motion found in this project, the reasons causing these alterations and their influences on muscles and ligaments should be further investigated in further study. These alterations of movement pattern should be also considered when designing or practicing this exercise program in clinics.

Table 7. Maximum joint angles in stance phase of stair decent

Joint angles	Normal		ACLD				ACLR			
			Noninvolved limb		Injured limb		Noninvolved limb		Injured limb	
	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)	Mean	(S.D.)
Hip joint										
flexion	15.50	(4.91)	20.77	(4.41) b	19.77	(3.80)	17.41	(6.34)	17.35	(4.51)
extension	-1.82	(5.07)	-4.79	(4.98)	-5.50	(6.55)	-1.50	(4.63)	-1.01	(5.48)
Int. rotation	3.19	(2.63)	3.65	(5.38)	4.32	(3.66)	3.27	(5.04)	5.96	(2.80)
Ext. rotation	7.18	(3.09)	9.68	(3.80)	7.64	(4.81) *	7.47	(3.30)	7.78	(2.78)
adduction	7.18	(2.28)	8.05	(2.44)	5.85	(2.23) *	9.18	(2.99)	6.11	(3.93) *
abduction	2.41	(1.94)	2.63	(3.00)	3.48	(3.53)	2.81	(2.83)	3.69	(2.66)
Knee joint										
flexion	76.72	(7.72)	81.49	(6.42)	72.97	(10.62) *	74.69	(3.99)	67.71	(10.15) a*
extension	-8.96	(5.39)	-8.33	(2.62)	-9.00	(6.09)	-7.73	(4.52)	-11.46	(4.21) *
Int. rotation	-1.34	(6.91)	2.29	(7.58)	-1.12	(6.30)	-0.44	(3.05)	-3.39	(4.94)
Ext. rotation	15.42	(8.67)	13.90	(9.43)	15.42	(6.05)	18.22	(5.95)	17.86	(6.08)
adduction	4.32	(5.21)	0.43	(3.79)	2.91	(6.01) *	-1.33	(3.19) b	0.42	(3.25) *
abduction	5.90	(3.41)	9.05	(4.38)	9.09	(7.42)	10.29	(5.32)	7.93	(4.02)
Ankle joint										
dorsiflexion	23.81	(4.56)	26.81	(4.80)	24.09	(5.61)	26.02	(7.21)	21.17	(6.65) *
plantarflexion	18.38	(5.53)	24.07	(4.21) b	20.69	(5.65) *	22.81	(5.68)	16.13	(2.46) *
adduction	7.66	(3.17)	9.21	(7.27)	6.38	(5.19)	9.69	(3.55)	7.47	(3.02)
abduction	4.78	(2.13)	4.49	(5.17)	5.75	(4.87)	3.92	(3.18)	4.92	(2.99)
inversion	8.71	(2.92)	9.39	(5.24)	10.58	(4.72)	8.85	(4.02)	9.29	(3.57)
eversion	7.80	(3.90)	8.06	(2.31)	6.48	(4.08)	6.22	(4.00)	6.82	(3.92)

* p-value<0.05 in comparing the injured with noninvolved limb in ACL groups with paired t-test.

[a] p-value<0.05 in comparing the injured limb in ACL groups with normal limb with ANOVA.

[b] p-value<0.05 in comparing the noninvolved limb in ACL groups with normal limb with ANOVA.

Table 8. Maximum joint moments in stance phase of stair decent

	Normal	ACLD		ACLR	
		Noninvolved liml	Injured limb	Noninvolved liml	Injured limb
	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)	Mean (S.D.)
Hip joint					
extensor	7.45% (2.07%)	7.13% (2.72%)	7.56% (2.07%)	7.16% (2.58%)	7.32% (2.57%)
flexor	4.60% (1.89%)	7.09% (2.37%) b	3.67% (2.10%) *	6.44% (1.15%) b	3.39% (1.50%) *
abductor	15.20% (2.33%)	14.73% (3.19%)	12.20% (2.60%) a*	14.40% (2.99%)	11.91% (1.66%) a*
adductor	1.08% (0.62%)	0.78% (1.14%)	1.04% (1.15%)	1.24% (0.98%)	1.24% (0.54%)
int. rotator	3.04% (0.80%)	2.69% (1.44%)	2.79% (1.29%)	2.50% (0.73%)	2.23% (0.61%)
ext. rotator	0.60% (0.41%)	0.73% (0.34%)	0.54% (0.46%)	0.56% (0.27%)	0.38% (0.23%)
Knee joint					
extensor	14.70% (2.28%)	17.88% (2.93%) b	11.08% (4.45%) a*	15.67% (3.02%)	9.91% (2.49%) a*
flexor	4.06% (1.12%)	3.94% (1.03%)	3.57% (1.44%)	3.83% (0.97%)	3.13% (0.77%) *
abductor	4.79% (1.69%)	4.62% (2.60%)	2.97% (1.58%) a*	4.00% (2.10%)	3.31% (1.54%)
adductor	1.56% (1.01%)	1.81% (1.69%)	2.56% (2.29%)	2.29% (1.39%)	1.65% (1.32%)
int. rotator	0.71% (0.57%)	0.58% (0.53%)	0.81% (0.72%)	0.49% (0.39%)	0.40% (0.32%)
ext. rotator	1.35% (0.61%)	1.18% (0.81%)	0.71% (0.32%) a*	1.10% (0.63%)	0.82% (0.37%)
Ankle joint					
plantarflexor	17.76% (2.14%)	16.06% (1.54%)	15.73% (2.32%) a	16.17% (1.70%)	14.40% (1.78%) a*
dorsiflexor	0.38% (1.04%)	0.11% (0.22%)	0.04% (0.14%)	0.00% (0.09%)	0.04% (0.16%)
evertor	1.26% (1.13%)	1.07% (1.31%)	1.14% (1.21%)	1.12% (0.74%)	0.54% (0.47%) *
invertor	2.21% (1.18%)	2.38% (1.47%)	1.69% (1.26%) *	1.37% (0.93%)	1.93% (0.58%)
abductor	0.21% (0.33%)	0.37% (0.78%)	0.26% (0.41%)	0.04% (0.07%)	0.05% (0.09%)
adductor	3.68% (1.07%)	3.39% (1.14%)	3.11% (1.21%)	3.26% (1.36%)	2.92% (0.94%)

* p-value<0.05 in comparing the injured with noninvolved limb in ACL groups with paired t-test.

[a] p-value<0.05 in comparing the injured limb in ACL groups with normal limb with ANOVA.

[b] p-value<0.05 in comparing the noninvolved limb in ACL groups with normal limb with ANOVA.

Conclusions

There was little difference in the kinematics and kinetics between ACL and normal subjects because the patients frequently performed the tested functional activities. They may have adapted their neuromusculoskeletal system to accomplish the tasks very close to normal. However, the asymmetries between the injured and noninvolved limbs were still noteworthy in ACL subjects. Agreed with past literature, quadriceps avoidance patterns, which revealed by reduced quadriceps activity and reduced knee extensor moments, occurred in ACL subjects during level walking, obstacle-crossing, and stair activities. Through detailed three-dimensional analysis of these functional activities, we found that the alterations of joint kinematics and kinetics in the involved limb occurred in all three planes, rather than in the sagittal plane, which was particularly emphasized in the past. It also indicated that neuromuscular or structural alterations existed in the ACL deficient and reconstructed knee during functional activities, such as reflex inhibition of muscle activation, muscle strength deficit, and dynamic instability. Because of the declined function at knee joint, lower extremity, as a linkage system, could develop a compensatory strategy for maintaining a stable trajectory of center of mass (COM) during activities. Therefore, a thorough investigation in the biomechanical and neuromuscular recovery in ACL subjects helps to understand the strategies of adaptation.

The influence of bracing on the joint kinematics, kinetics and the mean muscle activation was found to be insignificant during level walking. It indicated that knee braces do not affect muscle performance and activation, so muscle training is needed if proper compensation for the affected knee is required. Similar insignificant results were found in the joint kinematics and kinetic during stair activities. This suggests that bracing provides little effect during stair activities since the tibial anterior translation has been demonstrated to be very small during stair ascent [22]. Furthermore, the moderate knee flexion angle during stair activities could provide good stability at knee joint, since the quadriceps contraction generates smaller anterior draw force in this range [55,56]. In the future, more detail analysis should be included for the further discussion of the necessity of functional brace.

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國家衛生研究院九十一年度整合性醫藥衛生科技研究計畫

前十字韌帶損傷及重建後膝關節之生物力學與神經肌肉適應之分析

(Biomechanics and Neuromuscular Adaptation in ACL Patients)：

第三子計畫 － Neuromuscular Adaptation of Knee Muscles in Patients with ACL Deficiency

Annual Report 2002; Component Project 3

執行機構：高雄醫藥大學復健醫學系

子計畫主持人：蕭世芬

Responsible Investigator: Dr Shih-fen Hsiao, School of Rehabilitation Medicine, Kaohsiung Medical University

The Component Project 3 in 2002

In this year, Component Project 3 aimed to investigate the amount of muscle weakness due to ACL deficiency (ACL-D) bilaterally and the compensatory effects in the short and long term, with or without reconstruction (ACL-R). This years investigation were focused on the changes and recovery of the length : tension and force : velocity relationships of the knee flexors and extensors after ACL injury and reconstruction. The extent of voluntary failure to recruit motor units during maximal efforts was also investigated on both quadriceps and hamstrings using the twitch superimposition technique. Results of this component project should be able to explain the deficits and the adaptation after ACL injury and help the interpretation of the findings from other component projects. With the completion of this project, it is also hoped that the understanding of these neuromuscular and functional changes of knees would lead to a better treatment planning for patients with ACL-D and an optimized outcome. The result of this years investigation would help to establish a better structure of the rest of the experiments in the coming years.

Research Design and Methods:

Subject selection

Patients diagnosed with ACL deficiency and admitted for a reconstruction was recruited from the Departments of Orthopedic Surgery, of the Chung-Ho Memorial Hospital (Kaohsiung Medical University), Kaohsiung. The reconstruction procedure used was the bone-patella tendon-bone autograft. Consented subjects had a confirmed diagnosis of ACL injury without other knee pathology such as fracture or meniscus tears. For those who were < 18 years old, written consent was also obtained from their legal guardians. People who had previously injured their lower extremities and had to confine to bed for more than a month, previous history of neurological disorder such as stroke, those who are (or anticipated to be) pregnant, were excluded. Age-matched control subjects, also with written consent, were recruited locally for comparison.

Investigation protocol

This year's investigation was focused on the mechanical output of the knee muscles of ACL deficient knees. All patients were tested bilaterally in random order; experiments were scheduled before reconstruction, and then three and six months after the surgery. Though so far the sixth month test is yet to complete.

The control group was tested with the exact testing protocol except that they were only tested once on their preferred knee.

Methods:

All tests were performed with subjects sitting on an isokinetic dynamometer (KIN-COM

Auto-positioning, Chattanooga, USA), with the backrest positioned at 120° position while the knee hanged at the edge of the support (trunk/hip at slight flexion and knees at 90° flexion). The knee position varied as requested in different tests. The trunk, waist, thigh and shank were strapped, and the subjects were instructed to fold their arms in front of their chest to minimize unnecessary movement.

Force-velocity relationship were tested on both knee muscles, with a velocity spectrum of $50°\cdot sec^{-1}$ to $250°\cdot^{-1}$ of $50°\cdot sec^{-1}$ increments in randomized order. Subjects were asked to do three maximal voluntary contractions (MVCs) of the quadriceps and hamstrings at the range of 10° to 90° of flexion. Rest was given between contractions to minimize fatigue. Similarly, *length-tension relationship* of both quadriceps and hamstrings were examined within the same movement range; isometric MVCs of both knee muscles were performed in every 20° increments. Force, joint angle and velocity were recorded with build-in software for subsequent analysis. *Voluntary activation failure during isometric contraction* was detected by twitch superimposition technique using 1 Hz magnetic percutaneous nerve stimulation for both quadriceps and hamstrings (Magstim 200, Magstim, UK), with a 90 mm circular coil. The magnetic pulse was a biphasic triangular wave with 100 μs rise time, pulse width 1 ms, and delivered at 1 Hz or less. Both quadriceps and hamstrings were examined at 70° of knee flexion in the sitting position.

Results:

It was found to be difficult to recruit sufficient number of participants in a short time, as setting up the laboratory in KMU and combined possible resources took a long time. So far there have been five ACL-D participants received reconstruction; three of them had completed their second tests. There were also 15 volunteers serving as the control group. The details of the participants are listed in Table 1.

Table 1. Demographic data of the participants.

	ACL group (N=5)	Normal group (N=15)
Age (years)	24.00±3.90 (17-34)	22.93±0.80 (17-30)
Gender	2 males	11 males
Affected / tested side	1 right side	11 right side
Body weight (Kg)	59.40±4.52	67.47±2.60
Time between onset and operation (months)	33.2±14.8	-

The major findings in the first year are summarized here as Tables 2-7. Table 8 shows a brief result of those who came back for re-test.

Table 2. Force-velocity characteristics (*absolute* strength in Newton) of the knee muscles during isokinetic movement.

Quadriceps	Normal group (N=15)	Unaffected side, ACL group (N=5)	Affected side, ACL group (N=5)
$50°\cdot sec^{-1}$	622.1±49.6	577.8±49.4	393.0±126.0
$100°\cdot sec^{-1}$	530.7±38.7	479.2±41.8	418.2±74.6
$150°\cdot sec^{-1}$	443.2±36.6	432.4±51.1	358.6±54.9
$200°\cdot sec^{-1}$	388.1±37.0	370.0±49.2	320.0±57.1
$250°\cdot sec^{-1}$	372.8±28.8	352.8±35.5	332.2±65.1
Hamstrings			
$50°\cdot sec^{-1}$	377.3±33.3	248.2±52.0	226.7±51.8
$100°\cdot sec^{-1}$	331.3±29.3	247.0±50.1	206.8±53.2
$150°\cdot sec^{-1}$	295.6±29.0	224.4±44.9	189.0±40.9
$200°\cdot sec^{-1}$	276.5±29.6	236.6±50.5	195.0±42.2
$250°\cdot sec^{-1}$	247.9±24.8	207.8±44.8	154.3±46.8

Table 3. Force-velocity characteristics (*relative* strength in %, normalized by body weight) of the knee muscles during isokinetic movement.

Quadriceps	Normal group (N=15)	Unaffected side, ACL group (N=5)	Affected side, ACL group (N=5)
$50°\cdot sec^{-1}$	93.47±6.08	99.55±6.04	63.50±19.3
$100°\cdot sec^{-1}$	79.87±4.67	82.35±3.91	70.34±7.95
$150°\cdot sec^{-1}$	66.57±4.46	73.61±4.03	60.75±6.11
$200°\cdot sec^{-1}$	58.23±4.80	62.81±5.03	53.79±6.31
$250°\cdot sec^{-1}$	55.92±3.28	60.28±2.22	56.09±8.79
Hamstrings			
$50°\cdot sec^{-1}$	56.06±3.36	42.04±6.30	37.60±3.95
$100°\cdot sec^{-1}$	49.63±3.46	41.77±5.36	34.38±4.78
$150°\cdot sec^{-1}$	43.98±3.40	39.63±7.42	32.41±5.49
$200°\cdot sec^{-1}$	41.46±3.94	41.05±7.13	32.76±4.47
$250°\cdot sec^{-1}$	36.90±3.55	36.08±6.49	26.68±5.98

Table 4. Length-tension characteristics (*absolute* strength in Newton) of the knee muscles in isometric contractions

Quadriceps	Normal group (N=15)	Unaffected side, ACL group (N=5)	Affected side, ACL group (N=5)
10° of flexion	257.2±21.6	236.2±19.0	195.6±30.5
30° of flexion	387.4±31.6	335.2±17.9	263.8±37.4
50° of flexion	572.2±47.4	465.8±27.3	399.2±58.3
70° of flexion	682.5±60.9	608.8±71.5	496.8±52.9
90° of flexion	537.8±60.4	526.8±75.3	461.6±57.4
Hamstrings			
10° of flexion	290.5±23.7	237.0±14.3	201.8±28.0
30° of flexion	300.3±25.9	255.6±28.3	189.2±10.6
50° of flexion	292.5±24.7	259.0±29.4	195.2±24.4
70° of flexion	265.2±23.2	266.0±37.2	183.4±29.6
90° of flexion	208.1±26.5	222.4±37.7	178.8±34.6

Table 5. Length-tension characteristics (*relative* strength in %, normalized by body weight) of the knee muscles in isometric contractions

Quadriceps	Normal group (N=15)	Unaffected side, ACL group (N=5)	Affected side, ACL group (N=5)
10° of flexion	38.54±2.43	41.41±4.60	33.67±4.99
30° of flexion	58.40±3.92	58.04±2.06	45.07±4.99
50° of flexion	85.91±5.80	81.14±5.92	67.97±7.13
70° of flexion	101.93±6.96	105.1±11.4	85.15±5.84
90° of flexion	80.42±7.91	92.2±14.8	79.93±9.66
Hamstrings			
10° of flexion	43.65±2.95	41.33±3.27	34.98±5.05
30° of flexion	45.18±3.33	44.99±6.60	33.24±3.20
50° of flexion	44.02±3.18	45.41±6.52	33.69±3.93
70° of flexion	40.11±3.21	46.53±7.57	31.55±4.74
90° of flexion	31.25±3.51	38.71±7.16	30.55±5.23

Table 6. Angles where peak force occurred during isokinetic movement

Quadriceps	Normal group (N=15)	Unaffected side, ACL group (N=5)	Affected side, ACL group (N=5)
$50° \cdot sec^{-1}$	64.80±2.38	67.40±2.54	68.25±2.10
$100° \cdot sec^{-1}$	60.87±2.13	61.00±2.14	66.00±2.77
$150° \cdot sec^{-1}$	59.67±1.86	54.00±1.79	62.60±3.64
$200° \cdot sec^{-1}$	60.40±1.50	57.60±1.63	60.20±4.51
$250° \cdot sec^{-1}$	52.93±1.17	54.40±0.93	59.20±2.85
Hamstrings			
$50° \cdot sec^{-1}$	28.40±1.80	34.00±3.59	43.80±11.6
$100° \cdot sec^{-1}$	38.00±1.76	39.00±3.24	40.40±5.64
$150° \cdot sec^{-1}$	42.73±2.94	35.00±3.66	42.60±6.14
$200° \cdot sec^{-1}$	46.07±3.46	43.00±6.97	36.25±6.50
$250° \cdot sec^{-1}$	50.71±3.43	54.00±3.54	46.50±6.03

Table 7. Time-to-peak (in second) of the knee muscles in isometric contractions

Quadriceps	Normal group (N=15)	Unaffected side, ACL group (N=5)	Affected side, ACL group (N=5)
10° of flexion	0.831±0.113	0.868±0.080	0.638±0.057
30° of flexion	0.911±0.104	0.707±0.147	0.662±0.193
50° of flexion	0.919±0.116	0.738±0.203	0.612±0.026
70° of flexion	0.789±0.121	0.933±0.113	0.684±0.122
90° of flexion	0.831±0.101	0.670±0.264	0.552±0.073
Hamstrings			
10° of flexion	0.754±0.103	0.925±0.129	0.756±0.179
30° of flexion	0.796±0.101	0.918±0.095	0.606±0.115
50° of flexion	0.733±0.074	0.782±0.149	0.572±0.098
70° of flexion	0.778±0.089	0.750±0.069	0.640±0.095
90° of flexion	0.611±0.064	0.615±0.078	0.666±0.114

Table 8. Review at three months post operation with bone-patella tendon-bone technique (under the conditions that subjects do not receive regular physical therapy.

	Affected quads		Non-affected quads		Affected hams		Non-affected hams	
Force	Pre-OP	Post-3	Pre-OP	Post-3	Pre-OP	Post-3	Pre-OP	Post-3
0 degree /sec	524	379	684	822	249	241	309	360
50 degree /sec	493	290	603	748	243	346	282	391
250 degree /sec	388	173	362	411	193	272	277	317
Angle								
0 degree /sec	70	63	77	70	50	50	50	70
50 degree /sec	70	73	71	68	50	29	35	20
250 degree /sec	61	55	56	51	44	46	49	41
Activation failure	54%	33%	26%	19%	27%	24%	22%	4%

Overall in this project, the ACL deficient knees were weak in both flexion and extension movement; patients also seemed unable to move fast enough ($250°\cdot sec^{-1}$) as well as fatigue easily. There were also signs of activation failure in both knee muscles, bilaterally. For the unaffected side there were improvements after the reconstruction, though the affected quadriceps seemed to be even weaker in the retest, with the presence of some discomfort and incomplete range in extension. Therefore some issues should be examined further in order to clarify the effect of reconstruction and the recovery.

Conclusions of Year 2002

1. Muscle strength:

A. Both knee flexors and extensors were affected in the ACL deficient knee, with some insufficiency appeared in the contralateral extremity.

B. Despite the apparent weakness seen before operation, the affected quadriceps seemed to be even weaker at three months.

C. The hamstrings muscle at the affected side showed little changes at all testing velocities.

D. Both quadriceps and hamstrings at the unaffected side showed improvement.

E. There was evidence of extensor lag and sign of fatigue at the affected side. The length-tension relationship was altered for both quadriceps and hamstrings which need close monitoring in the recovery course.

2. Activation failure:

A. Activation failure was evident before reconstruction and three months after (with some improvement), in all muscle bilaterally.

B. Among all, the unaffected hamstrings seemed to regain the activation ability best.

C. The affected quadriceps showed some improvement of activation though much weaker muscle. It is then suggest that at this stage (three months post op) muscle atrophy (loss of muscle fibres) rather than inability to recruit motor units was the cause of weakness. Further follow-up in the sixth month is necessary.