身心障礙者行動輔具之研發與人員之培訓(2/2)－長腿支架功能最佳化設計

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主持人：呂東武　　台灣大學醫學工程學研究所

☑ 出席國際學術會議心得報告及發表之論文各一份

一、中文摘要

長腿支架（髖膝踝足矯具）係脊髓損傷(Spinal Cord Injury, SCI)病患最常用的行動輔具之一。以往的支架設計常限制下肢多數關節的活動度以保持步行之穩定性，因而導致人體重心側向位移過大，不僅費力且能量耗損亦大。本計劃針對此一缺失尋求改良之道。首先建立一套人體三維動力學模型，並利用最佳控制理論(Optimal Control Theory)，透過改變該模型的系統拘束條件，模擬穿戴不同輔具之步態行為及其動力特性，再應用最佳設計理論(Optimal Design Theory)進行長腿支架實作設計，求得使系統耗能最小之最佳設計。

本二年期計劃於第一年完成建立人體三維模型，並透過理論分析與模擬評估，證實髖膝關節連動式設計之長腿支架確實可改善現行長腿支架耗能過大之現象。本計劃第二年除針對第一年研究成果作更深入探討外，進一步根據分析結果利用電腦輔助設計與製造(CAD/CAM)技術進行新式支架之細部設計，並選定受試者進行該支架的原型品製作。新式支架之設計理念在於儘可能地恢復病人之正常步態以使耗能最小。根據正常步態週期中髖膝關節運動學關係，使新式支架之膝關節於矢狀面上相對應於髖關節作連動式運動，讓病人於步態擺盪期有足夠地板間隙(floor clearance)，以改善傳統支架為了達到足夠間隙而使重心偏移過大以致耗能過度增加之情形。

關鍵詞：長腿支架、電腦模擬、最佳控制、最佳設計、步態分析、關節力矩。

二、前言

正常步態係藉由人體各關節間的協調機制減少人體重心的上下及左右位移量，以達到耗能最小之目的。長腿支架雖廣為脊髓損傷病患採用以輔助其行動，然而傳統設計為了達到人體行動之穩定性，均限制膝關節與踝關節之屈曲作用，因而造成穿戴者於步行時因重心起伏過大而使能量損耗大增[1]。因此本二年期計劃主要目的即在尋求一降低步行耗能之長腿支架最佳設計，以補傳統設計之不足。

在設計適合截癱病患使用之長腿支架時，人體數學模型的建立將有助於模擬穿戴輔具前後之步態行為，及後續輔具設計之功能評估。Tashman [5]曾針對一穿戴 RGO(Reciprocating Gait Orthosis)之病患建立一三維人體模型，並以該模型探討該病患於擺盪期之特性。惟該模型僅限於單一病患，不能廣用於探討正常人或使用不同支架之不同病患的步態行為。本計劃發展之三維人體模型係一直接動力學模型，能允許系統拘束條件之改變，故可針對不同病患、不同支架之設計，彈性調整人體模型中各自由度之拘束條件，以模擬人體穿戴 RGO 或不同支架之步態過程及穿戴效果。為改善現有支架設計較正常步態有過度耗能之情形，本計劃利用正常步態之髖膝關節間的運動學關係，設計一耦合連桿，允許膝關節相對應於髖關節做屈曲角度運動，改善現行 RGO 關節活動度不足之缺陷。並透過上述人體電腦模型之模擬分析，比較正常、穿戴現行 RGO 及新型支架三者之步態歷時耗能，驗證新設計的理念。

本計劃除了從理論研究出發，發展人體電腦模型外，更著重於長腿支架之實作設計。經理論評估新式設計的可行性後，即利用最佳設計理論尋求使系統耗能最小之最佳設計。本計劃第一年之研究證實了新式設計的確明顯改善既有設計之耗能，第二年則根據此一理念進行新式支架的細部最佳設計，並在穩定度與活動度之配合上完成長腿支架功能性的改良。

三、研究方法

本計劃三維人體模型之建立，係將人體模擬成由模型關節連接的 8 個剛體肢段，如圖一所示。其中人體軀幹(torso)相對於骨盆(pelvis)及骨盆相對於大腿(thigh)間係以球關節模擬。大腿相對於小腿(shank)及小腿相對於足部(foot)之運動則以鉸接模擬。地面與足部間之相對運動則以三個方向之滑動拘束式作考量。透過相關人體測量學參數之適當選取，該模型可用於模擬任一病患。

人體三維模型運動方程式之建立，係利用 Kane's Equation 經降階推導後求得，如下：

$$\dot{q} = f(q, u, t) \quad (1)$$

其中 q 為系統之狀態變數(state variable)，即系統之運動自由度，u 為系統之控制變數(control variable)，即系統所受之關節力矩。t 為時間。

此人體三維模型在正常步態下共有 19 個系統自由度。模擬人體穿戴 RGO 之步態行為係於髖關節施加兩個拘束條件式，並各加一拘束條件式於膝關節及踝關節。換言之，僅允許下肢在髖關節矢狀面上進行屈伸運動，而膝及踝關節則不具任何運動，使系統自由度減為 11 個。如此模擬之結果將使人體下肢成不自然之單擺運動，為保持足部與地面之足夠間隙，必導致身體重心位置起伏變化過大，而消耗過多能量。為改良此一缺陷，本計畫在模擬人體穿戴新式設計之長腿支架的步態行為上，利用了正常步態週期中髖關節與膝關節於矢狀面上的角度變化情形，建立一髖膝關節屈曲角度間的數學函式關係，允許膝關節依實際髖關節運動而有相對應之角度變化。故此新式設計之系統自由度為 13 個。

步態模擬過程中，本計畫應用最佳控制方法分析各式支架之力動學行為，其目標函數為：

$$min \quad J = \int_{t_o}^{t_f} \sum_{i=1}^{N} w_i(q_{pred} - q_{obs})dt \quad (2)$$

其中 t_o 與 t_f 分別為初始與結束時間，N 為系統自由度個數，q_{pred} 為數值預測結果，q_{obs} 為實驗觀測值，w_i 為系統各自由度之誤差加權值。由於初始猜值(initial guess)的選定直接影響最佳化的搜尋過程及收斂效果，本計劃以逆向動力學計算所得之關節力矩作為初始猜值，以避免電腦不必要之冗長數值運算。

為簡化計算分析，可藉由控制變數於時域中的離散化過程，將上述式(2)之非線性最佳控制問題轉換為非線性規劃問題，並以 Sequential Quadratic Programming (SQP)最佳化方法求解之[6]。利用式(2)求得之模型中各關節力矩與角度等力動學與運動學資料，可進一步計算步態過程中的能量變化，以瞭解各式支架設計之耗能情形。

本計畫新式長腿支架的設計旨在改良並整合傳統 RGO。傳統 RGO 限制了人體下肢系統之膝關節與踝關節的運動，僅允許髖關節做屈曲，其左右髖關節作反對稱屈伸運動，亦即當左髖關節向前屈曲 5 度時，右髖關節則同步向後伸展 5 度。新式支架除了採用 RGO 左右髖關節的連動式設計外，更放寬了膝關節的限制，允許膝關節對髖關節的屈曲角度變化作相對應運動，以改善步態過程中，RGO 限制膝關節屈曲運動所導致的重心起伏過大帶來損耗過多能量。其中，髖關節與膝關節於整個步態週期的角度關係，是利用正常步行過程中髖膝關節角度變化的對應結果，如圖二、三所示。

正常步態週期中髖關節與膝關節屈曲角度的歷時關係如圖二所示。A、B 兩點分別為髖關節於整個步態週期中的最大伸展角度與最大屈曲角度，C 點為腳跟著地時之角度位置。圖三之虛線部分（含 A 點至 C 點順時針方向之實線部分）係分別以圖二之髖關節與膝關節屈曲角度作橫座標與縱座標所得之髖膝關節角度對應關係圖。在實際設計上，為穩定性考量，將 C 點至 A 點的站立期曲線關係改以圖中實線部分替代。亦即於站立期間，髖關節屈曲時所對應之膝關節角度為零，以確保步行過程中膝關節之穩定性。透過此一新式支架之髖膝關節連動式設計，病患可利用髖關節主動之運作或以身體擺盪方式來使髖關節作屈伸運動，並以圖三之髖膝關係帶動膝關節作相對應運動，以達到耗能最小之目的。

新式長腿支架的設計實作，在儘量不影響使用者之行動能力與降低步行過程之能量耗損等多重考量下，本計畫應用最佳設計理論，假設該支架之材料密度為已知參數，以新式支架總質量最小作為目標函數，依選定受試者的人體參數學資料，求得支架各部最佳尺寸設計與配置。再運用電腦輔助設計軟體(AutoCAD R14, Autodesk, U.S.A)，將此一設計結果的力學機構關係轉換為電腦程式碼(CAM codes)，利用 CNC 沖床技術製作原型品之各部結構，並針對各部關節間的吻合度與精確度作仔細的組裝與配置。於原型品完成後，更作進一步的校正與測試工作，以滿足受試病患對該新式支架在使用上的舒適度。

四、結果與討論

最佳控制法中目標函數內之狀態變數與控制變數間係以運動方程式互為關聯。目標函數值不僅受控制變數給定之影響亦與起始條件(Initial Condition)有關，其最佳化解之搜尋過程即視為一控制過程。系統模型為受控之個體，運動方程式則規範該個體受控之反應。控制變數為主動之變數，其不僅直接影響目標函數值的大小，亦透過運動方程式來改變狀態變數而間接影響目標函數。目標函數則為系統之回饋，用以決定控制之方向。此外，人體各關節之運動範圍則以狀態變數對應該關節之限制條件表示之。當目標函數達到極值時，表示所控制之結果為所求。該法中不僅可藉由觀察已知之實驗數據控制系統，藉以驗證模型之正確性外，並可進一步改變控制回饋以預測模型之反應。此外，以直接動力學方法求解可驗證系統運動方程式的正確性，而以逆向動力學求解則可作為最佳控制法的初始猜值。兩種方法亦對模型之正確性提供佐證。

圖四為正常、穿戴現行 RGO 及新式設計支架之能量歷時圖。由圖中得知，步行過程中新式設計之支架確實使位能的擾動程度減少，降低了人體不必要的能量消耗。其主要理念則來自於步態過程中髖關節及膝關節間的協調機制。

於長腿支架之最佳化設計製作中，本計畫將髖膝關節之連動式設計理念納入考量，且為避免新式支架之額外質量造成過多能量耗損，亦考慮支架總質量為最小。圖五至八為針對一特定受試病患之新式長腿支架的最佳設計結果。其中，圖五為全組立圖，係新式支架的總體設計圖。圖六至八則為主要部分之細部設計圖，用以說明髖膝關節間之細部構造與運作機制。圖九則為此新式支架之三維立體圖。茲將主要構造設計原理說明如下：。

在髖膝關節間的連動式設計部分，如圖五所示，本計畫係以兩條直徑 2mm 之輕質鋼索處理，以降低受試病患之荷重負擔。而髖膝關節間的角度運動係以圖三之關係作對應變化。於站立期間(stance phase)，髖膝關節的角度位置將以圖三實曲線部分之C點至A點的順時鐘方向作變化。於擺盪期間(swing phase)，則由 A 點至 C 點的順時方向角度關係作運動。此外，新式支架設計亦允許使用者自行調整髖關節角度的屈曲伸展範圍，為病患提供人性化的彈性設計。

圖六為腰部轉盤組立圖，圖中轉盤與鋼索轉盤係用以控制接往小腿支架上之兩條鋼索，藉以達成膝關節相對應於髖關節作屈曲伸展運動。此外，透過轉盤上之連桿帶動，可間接經由腰部支架上的背部曲柄來控制左右髖關節的相對運動。而轉盤上之螺絲係作為控制髖關節角度運動的範圍。置於轉盤與鋼索轉盤間的扭力彈簧則用來控制髖膝關節間的角度關係。圖七為膝部結構組立圖，利用圖中之壓桿設計來控制膝部關節運動的開閉。當受試病患之足部著地時，壓桿將帶動擋塊牽動大腿支架與小腿支架間的膝部 LOCK 鍵來鎖住膝關節運動，用以避免病患於站立時因膝關節運動所帶來的不穩定性。圖八則為大腿支架釋放組立圖，利用大腿 LOCK 鍵之閉開動作，來固定或解除髖關節之角度位置，前者將使髖膝關節之連桿機制發生作用，後者則釋放之。此外，新式支架設計仍保持踝關節角度為固定，故未來可對此部分再作進一步改良。

五、結論

傳統支架因以病患的自身穩定性作為設計上之優先考量，導致步行過程中能量消耗過大。為改善此耗能現象，本計劃於第一年建立三維人體模型，首先提供了一個分析評估各式支架對人體步態影響的極佳工具。並藉由該模型驗證了關節連動式設計比傳統支架更能充分考慮步行過程之穩定性與降低耗能。第二年則進行支架本體之細部設計，由選定之受試者製作原型品，並進行臨床評估分析。所得結果亦與計畫第一年的研究成果相互呼應，驗證了本計畫之設計理念。

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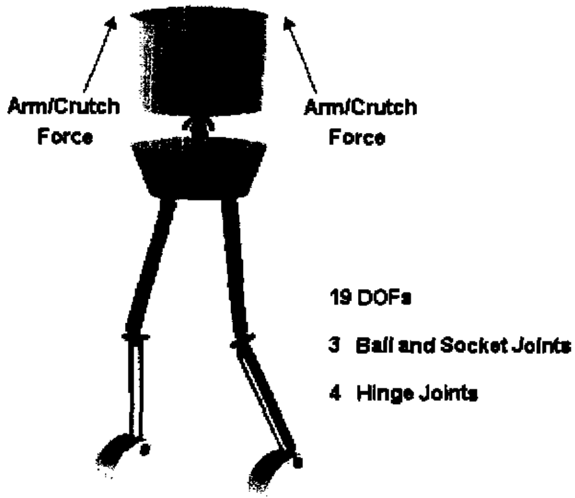
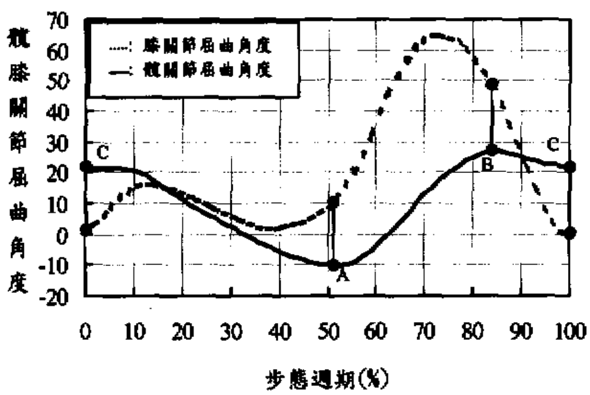
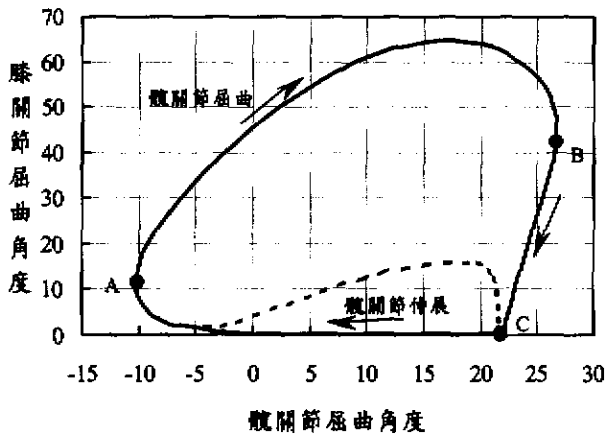


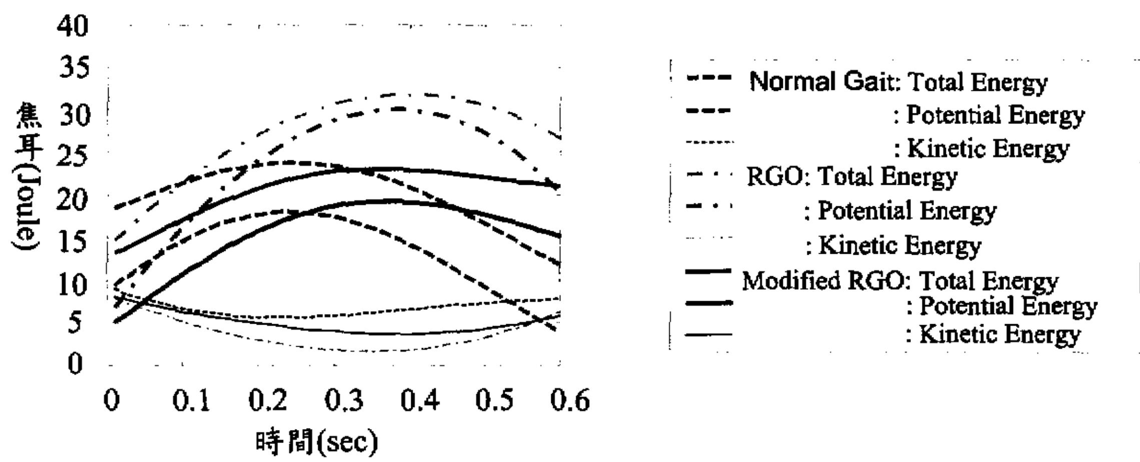
圖一 三維人體模型圖



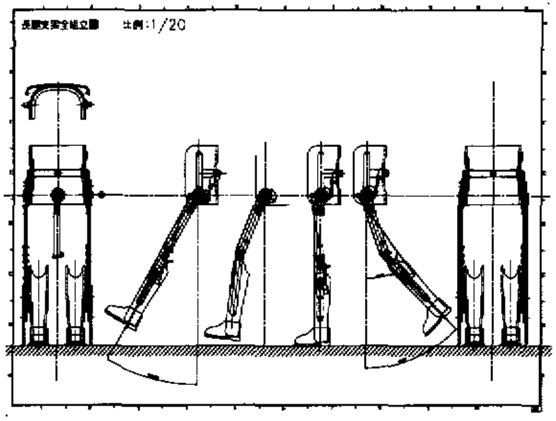
圖二 步態週期中髖膝關節屈曲角度運動學



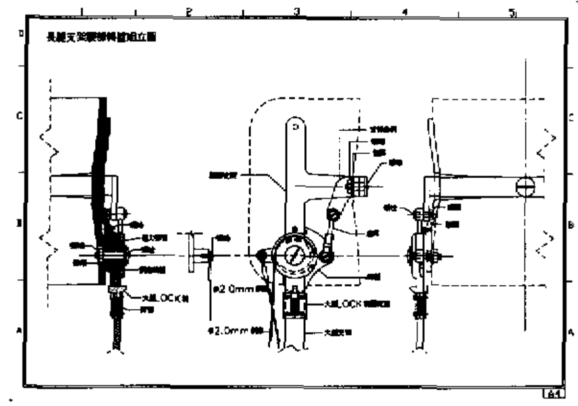
圖三 髖膝關節屈曲角度關係圖。圖中虛線部分（含 A-B-C 之曲線部分）為正常步態週期之實際髖膝關節角度關係；實線部分為實際模擬之結果。



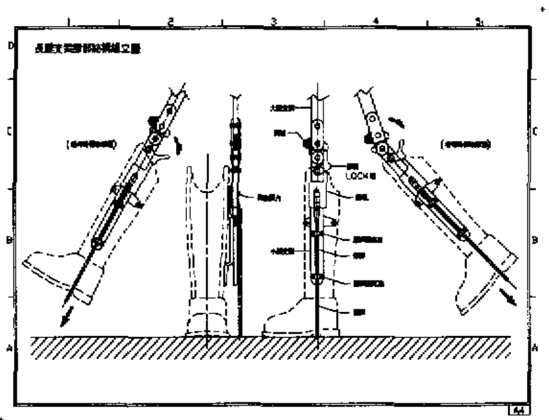
圖四 能量歷時圖



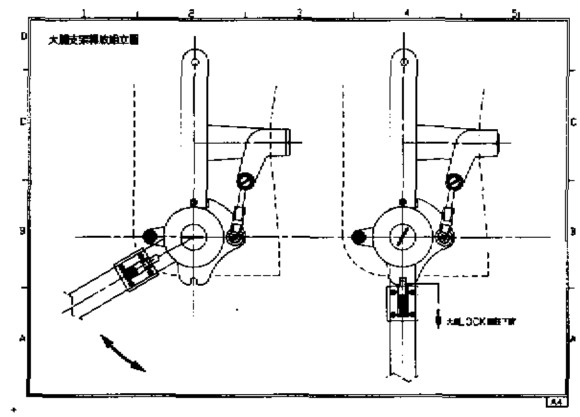
圖五 長腿支架全組立圖



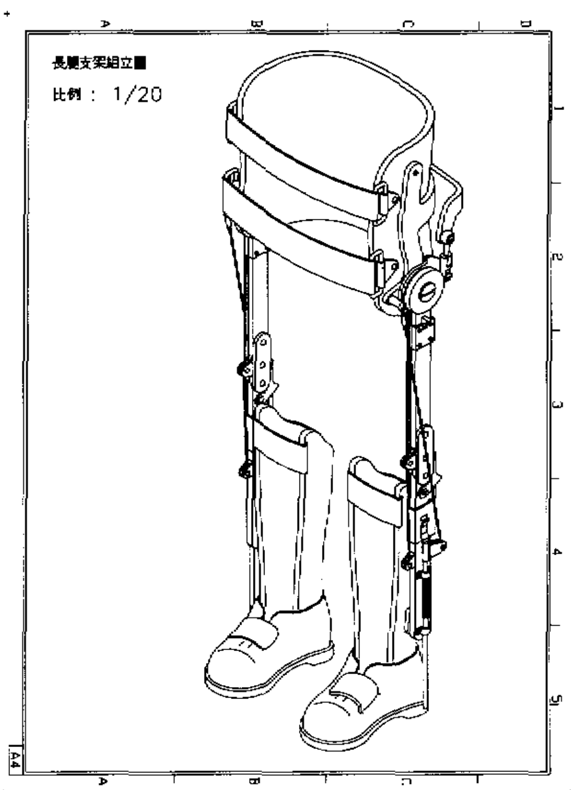
圖六 長腿支架腰部轉盤組立圖



圖七 長腿支架膝部結構組立圖



圖八 長腿支架大腿支架釋放組立圖



圖九 長腿支架三維立體圖

出國心得報告

呂東武

中華民國九十年七月二十五日

此次出國係爲了參加於瑞士 Swiss Institute of Technology Zurich 舉行之國際生物力學學會第十八屆大會，另有四位研究生隨行參與。個人在會議中發表了兩篇學術論文，題目分別爲"Influence of Functional Braces on the Kinetics of Anterior Cruciate Ligament-Injured Knees During Walking"及"*In Vivo* Measurement of Three-Dimensional Scapular Kinematics Using Stereophotogrammetry with Artificial Neural Networks"。

個人於七月五日搭乘瑞亞航空班機飛往 Zurich，於七月六日抵達。七月八日至十三日全程參與會議，參加會議人數約有三千人。個人論文發表時間爲七月九日下午三時三十分及下午五時，過程順利且反映熱烈。於會議後續幾日中，陸續有各國研究者前來接觸討論，其中有來自英國牛津大學、美國哈佛大學、荷蘭 Delft 大學、加拿大 Calgary 大學、義大利羅馬大學等相關領域之教授及研究者。結交不少學術界朋友，獲得不少新知，也爲後續跨國合作計劃建立初步共識。會議中並邀請數位生物力學領域之著名學者演講，包括 Prof. D.A. Winter、Prof. K. An、Prof. B.M. Nigg 等，聆聽他們的演講受益良多。會議另安排半天行程參加 Oxford Metrics 提供的 Polygon 及 Body Builder 訓練課程，亦獲益匪淺。七月十四日參觀主辦單位 ETH Zurich 生物力學實驗室後，於等待十八日搭乘班機返國期間進行私人訪問行程，對瑞士與 Zurich 因此有較進一步了解。

整體而言，此次出國參加國際學術會議，可謂收穫良多。不只將自己的研究成果與各國學者分享，更吸收不少新知。而隨行學生亦藉由此一會議對相關研究領域有更進一步認識，也增廣了許多見聞。

In Vivo Measurement of Three-Dimensional Scapular Kinematics Using Stereophotogrammetry with Artificial Neural Networks

Lu, T.-W., Li, G.-J., Kuo, M.-Y.,* Chang, L.-Y.* and Hsu, H.-C.*+
Institute of Biomedical Engineering, National Taiwan University, Taipei 100, Taiwan
*School of Physical Therapy, China Medical College, Taichung 400, Taiwan
+Department of Orthopedics, China Medical College Hospital, Taichung 400, Taiwan

Introduction

Knowledge of the scapular kinematics is essential for the assessment of shoulder function and the study of the biomechanics of the shoulder complex, contributing to the understanding of the aetiology and treatment of relevant joint diseases. Due to the large relative movement between the scapula and the overlying skin, the *in vivo* measurement of the scapular motion has been made only at static positions (e.g. Pronk, 1992; Johnson et al., 1993) unless invasive methods (Karduna et al., 1999) are used. Because of the difficulty, the kinematics of the shoulder complex has usually been described by the motion of the humerus relative to the trunk (Davis et al., 1997; Newsam et al., 1999). A method for the estimation of the scapular kinematics during dynamic movement is necessary if individual joints of the complex, such as the glenohumeral joint, are to be studied. A new method based on artificial neural networks (ANN) was developed for noninvasive *in vivo* estimation of the scapular kinematics during dynamic activities. Markers attached to scapular bone-pins were used to measure the true motion of the scapula for further comparison.

Materials and Methods

Four male volunteers performed static and dynamic tests in a gait lab equipped with a 7-camera motion analysis system (Vicon 370, Oxford Metrics, U.K.). The static tests involved arm elevation at discrete positions in the sagittal, frontal and scapular planes as well as scapular elevation/depression and medial/lateral displacement around the chest wall. A locator with 3 markers indicated the scapular position while those of the trunk and humerus were each described with four skin markers. Dynamic tests included functional activities involving continuous arm elevation such as forward reaching, head touching and wheelchair propulsion, during which only movements of the trunk and humerus were monitored. During all the tests, four markers attached to scapular bone-pins were used to provide the true positions of the scapula. Movements of the humerus and scapula were described relative to the trunk. The rotational movements of the humerus were defined using an Euler angle sequence consisting of the plane of elevation, amount of elevation and internal/external rotation (y-z'-y''). Rotation sequence for the scapula was internal/external rotation, upward/downward rotation and posterior/anterior tilting (y-z' -x'').

For each subject, a three-layered, fully connected, feed-forward, back-propagation network with 20 hidden neurons was constructed and trained with the humeral (input) and scapular (targets) static data (both rotational and translational components), Figure 1. The ANN was then used to estimate scapular positions from humeral data during dynamic tests. The estimated scapular motion was compared to those obtained with skeletal markers.

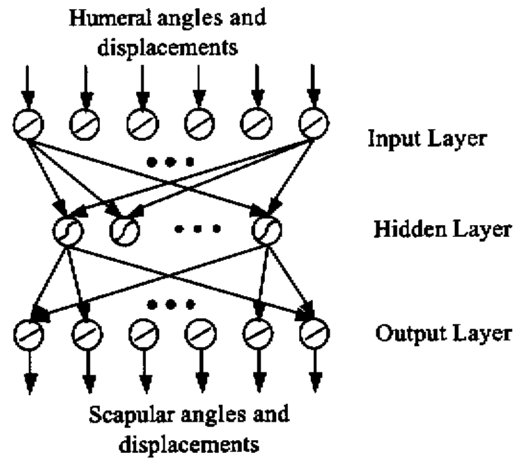


Figure 1. The artificial neural network describing the relation between the humeral angles and displacements with those of the scapula.

Results

An ANN was successfully trained to produce outputs closely matched the targets for each subject. Good agreement between the true and estimated movement patterns was found during wheelchair propulsion with RMS errors of 3.9, 1.9 and 2.1 degrees in protraction, lateral rotation and tilt respectively, Figure 2. Corresponding figures for forward reaching and touching head were (2.0, 2.4, 2.9) and (2.0, 3.1, 2.6) respectively.

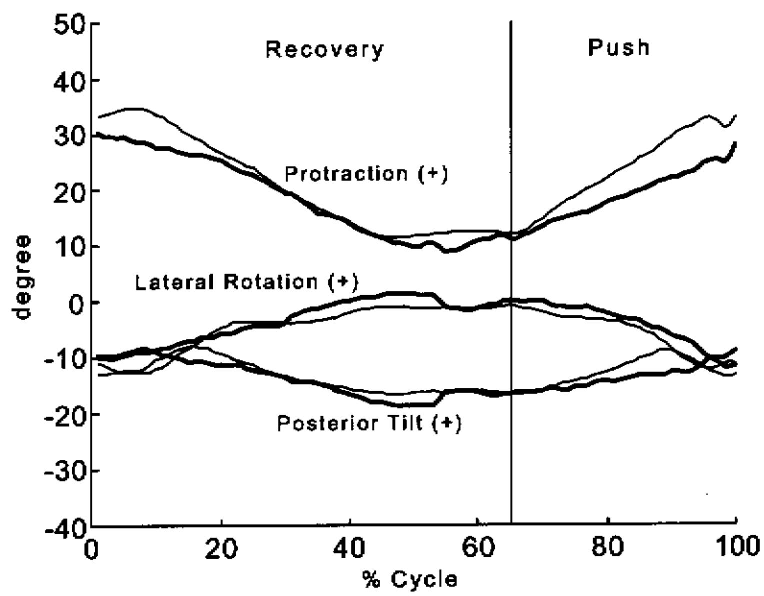


Figure 2. Typical ANN-estimated (thin) and true (thick) scapular motion patterns during wheelchair propulsion.

Discussion

Although static tests may reveal some features of the scapulohumeral motion during simple arm elevation, for a 3D activity it is difficult to define a series of static positions that follow the trajectory of the movement (van der Helm and Veeger, 1996). The ANN-based method was shown to be a promising method for the determination of the scapular kinematics during dynamic activities. It is suggested that the method will help extend the application of skin marker-based motion analysis systems to the studies of the musculoskeletal mechanics of the shoulder complex and upper extremities during dynamic movements.

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Influence of Functional Braces on the Kinetics of Anterior Cruciate Ligament-Injured Knees During Walking

Tung-Wu Lu, Yi-Cheng Hsieh, Gen-Jia Li, Hsiu-Chen Lin*, Horng-Chang Hsu+
Institute of Biomedical Engineering, National Taiwan University, Taipei 100, Taiwan
*School of Physical Therapy, China Medical College, Taichung 400, Taiwan
+Department of Orthopedics, China Medical College Hospital, Taichung 400, Taiwan

Introduction

Individuals with anterior cruciate ligament (ACL) injuries have been suggested to avoid quadriceps contraction with reduced extension moments during stance phase of gait to prevent excessive anterior tibial displacement (Andriacchi et al., 1993). Functional knee braces have been designed to provide necessary anterior/posterior stability (DeVita et al., 1997, DeVita et al., 1996) for these individuals. Therefore, studies on the kinetics of the knee with bracing have been on the changes of gait components in the sagittal plane, such as flexion and extension moments. Knowledge of their influence on the kinetics of the knee has not been available though with clinical significance. In the present study, a model of the lower limb combined with a specific marker system was developed and used to study the mechanics of the knee joint in individuals with ACL injuries without and with braces during gait.

Method

Ten ACL-injured subjects (mean age: 24.1, height: 168.9cm, weight: 66.1kg) were each fitted with suitable DonJoy Goldpoint braces (Smith & Nephew DonJoy Inc.) and walked at self-selected pace first without and then with braces in a gait laboratory equipped with a 7-camera motion analysis system (Vicon, Oxford Metrics, UK) and two force plates (AMTI, USA). A marker system was developed to enable the measurement of the bony landmarks around the knee joint while with braces. At least 3 successful trials for each condition were collected. A model of the lower limb was developed and used with inverse dynamics approach to calculate the forces and moments at the joints. Anthropometric data for the subjects were estimated using Dempster's coefficients (Winter, 1990). Note that the model knee joint allowed translation on the tibial plateau, an important consideration for ACL-injured patients. Angular impulses at the knee in three planes were calculated as the areas enclosed by the moment curves and x axes. Peak values for each moment component were also obtained, Figure 1. Comparisons of the angular impulses and peak moments between the bracing conditions and between injured and normal knees were made using paired samples t-test with a significance level of 0.05

Results & Discussion

Without bracing, angular impulses and peak moments at the injured knees were smaller than at the normal knees ($p<0.05$). With braces peak adduction and internal rotation moments and impulses at the injured knees increased significantly ($p<0.05$) while those at the normal sides remained relatively unchanged, Figure 2. No significant increase in flexion impulses and peak flexion moments was found at the injured and normal knees with bracing. With bracing, impulses and peak moments for adduction and internal rotation at the injured knees were not significantly different from the normal ones while angular flexion impulses at the injured knees remained significantly lower than the normal sides.

Functional braces were shown to increase significantly the peak internal rotation and adduction moments and angular impulses at the injured knees, improving the bilateral kinetic symmetry in the frontal and transverse planes. Although knee braces were designed to improve sagittal plane stability for ACL-injured patients, no significant improvement was found in this plane in terms of rotational stability for the subjects studied. The present study suggests that knee braces may not be effective in increasing sagittal plane stability, both rotational and translational, in ACL-injured patients.

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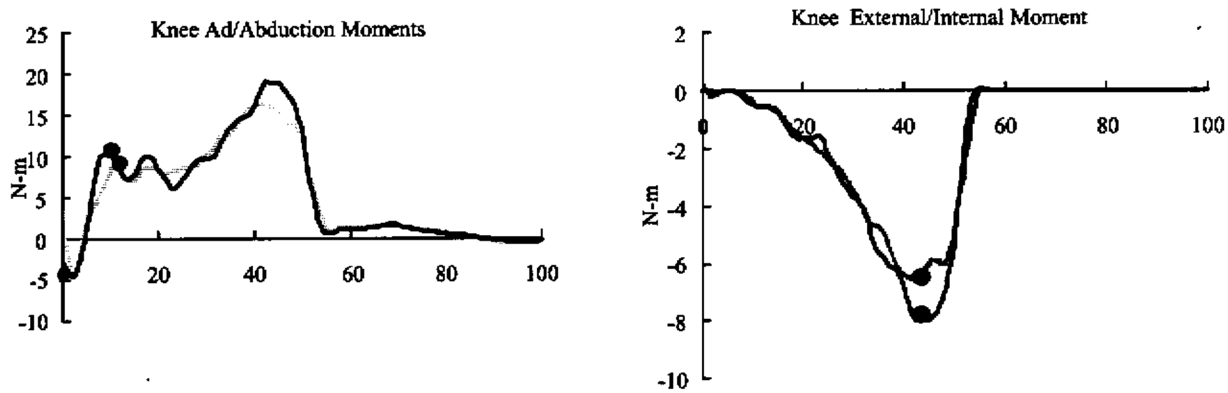

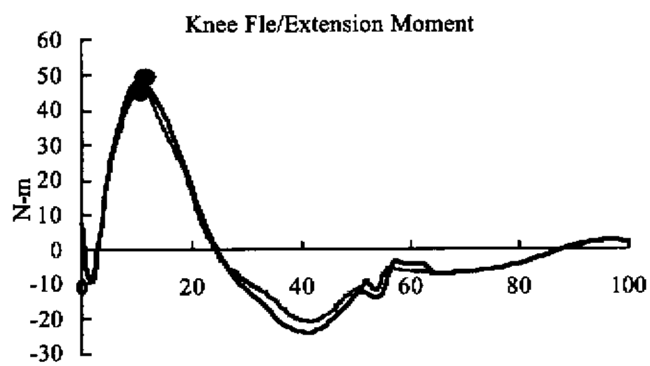

Figure 1: Knee moment curves from a typical subject during a gait cycle. The thick black lines are the moments for the with-brace conditions and the gray lines without-brace. The solid circles indicate the peak moments selected for statistical analysis.

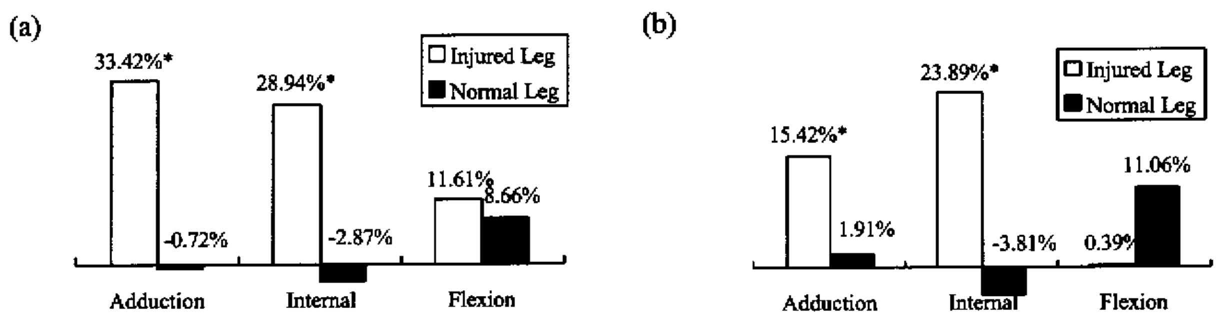

Figure 2. Average changes of (a) the peak moments and (b) angular impulses at the knee after wearing braces. (* : $p<0.05$)