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足部縦アーチの荷重による変化の検討：CT による３次元解析

Analysis of longitudinal arch of the foot in weightbearing and nonweightbearing using 3-dimensional computed tomography

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Key words：足（foot），縦アーチ（longitudinal arch），CT 画像（Computed tomography），３次元解析（three-dimensional analysis），荷重（weightbearing）

要　旨

足縦アーチの荷重による変化を，荷重・非荷重時の足部CTを３次元モデル化することで評価した．対象は足部疾患の既往のない男性12例12足であった．軸荷重装置を使用し，非荷重時は2kg，荷重時は体重の1/3を片脚に負荷しCT撮影した．画像解析ソフトを用いて荷重時と非荷重時の３次元足モデルを舟状骨で重ね合わせ，arch height index（以下 AHI）を内外側縦アーチについてそれぞれ算出し比較検討した．内側縦アーチの AHI は非荷重時に比べ荷重時に有意に低下したが，外側縦アーチの AHI は非荷重時，荷重時で有意差を認めなかった．本研究結果から，足縦アーチは荷重に対して内側と外側とで異なる動態を示すことが示された．

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緒　言

足は３つのアーチ構造（内側縦アーチ，外側縦アーチ，横アーチ）を有する．縦アーチ構造は歩行時にばねとして働き，蹴りだし力を高めて歩行の効率を良くし，体重移動を円滑にすると言われている[1]．

画像による縦アーチの評価法としては，X 線写真，足底圧，フットプリントなどの報告がある．しかし，これらの方法は足アーチの３次元構造を評価するには限界があり，生体内の荷重によるアライメント変化を正確に評価する方法はいまだ確立されていない．特に，足縦アーチを構成する内側と外側縦アーチの構造を個別に可視化し，それぞれのアーチにおける荷重の影響を研究した報告はない．

筆者らは，以前から荷重条件下に撮影したCTを用いた足アライメント評価方法を研究してきた．

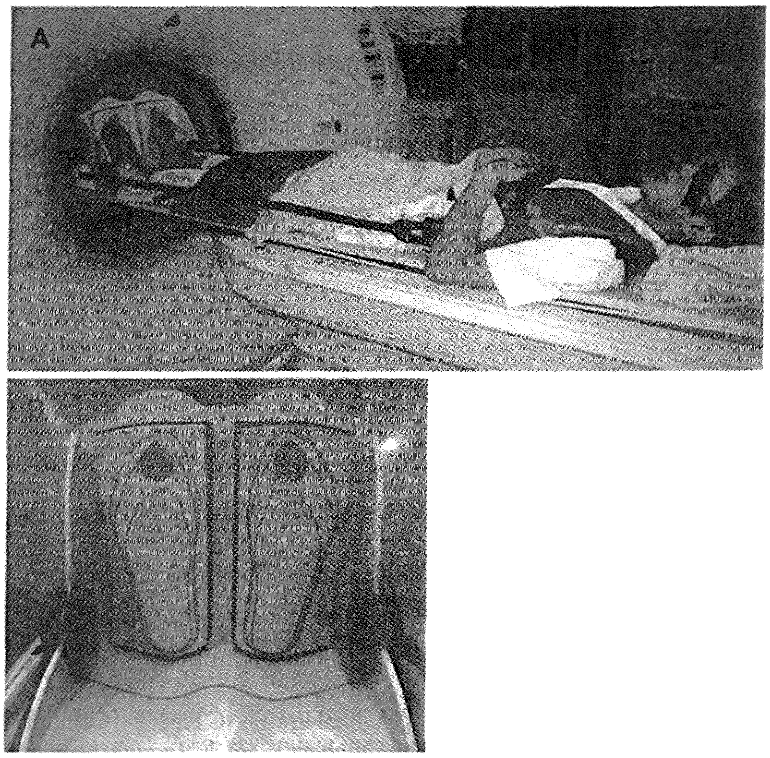

図1. 軸荷重装置を用いたCT撮影法
A：体幹に装着したベストと足台とをつなぐバンドを縮めていくことで，仰臥位でも足部に体重を模した荷重が負荷される．
B：足台の足接地面．足型がプリントされており，撮影肢位の再現性を向上させている．足台には片足ごとに体重計が内蔵されており，加わった荷重が計測可能である．

足の横アーチについては，荷重・非荷重時の足部CT3次元モデルを重ね合わせることで横アーチの動態を評価した[2]．この方法により，従来の計測方法では困難であった横アーチの各レベル（中足骨頭レベル，中足骨基部レベル，楔状骨一立方骨レベル）ごとの荷重による構造的変化を評価することが可能となった．

本研究では，荷重・非荷重条件での足部CT3次元モデルを用いて内側と外側縦アーチの荷重による変化を可視化し，定量的に比較検討することを試みた．

対象と方法

対象は足部外傷・障害の既往のない男性12例12足で，平均年齢は29.3歳(26-41歳)であった．CT撮影肢位は仰臥位で足関節中間位とした．CT撮影は脛骨遠位から足趾先端までの0.5mmスライスで行った．荷重には軸荷重装置(L-Spine，Dynawell社)を使用し，非荷重時として2kg，荷重時として体重の1/3を片脚に負荷した．荷重時には両脚で体重の2/3が負荷された状態で，両足を同時にCT撮影した．(図1)

CT撮影で得られた画像データから，骨表面の輪郭を抽出することにより，骨表面3次元モデルを作製した．(図2)次に画像解析ソフト(3-matic5.01®，Materialise社)を用い，荷重時と非荷重時モデルの舟状骨を完全に重ね合わせた．内外側の縦アーチ評価のため，第1趾列レベルを内側縦アーチ，

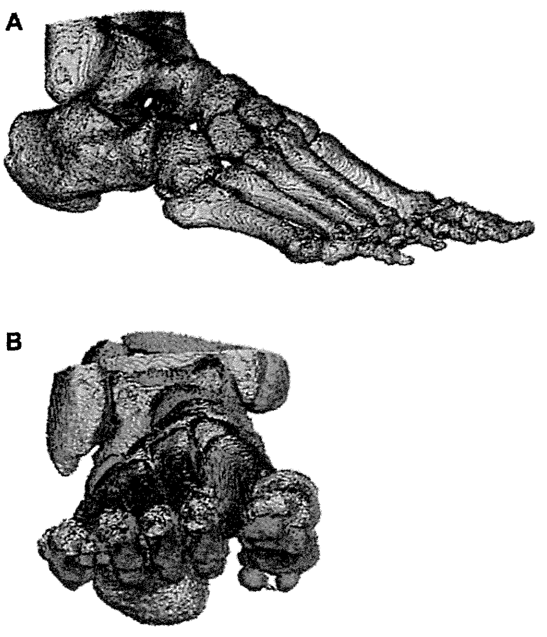

図2. 足の骨表面3次元モデル
A：非荷重時
B：荷重・非荷重条件の重ね合わせモデル
（黄：荷重時，黒：非荷重時）

第5趾列レベルを外側縦アーチとし，それぞれの断面図を作製した．（図3）各レベルにおける断面の決定方法は以下のとおりである．1）第1趾列レベル：第1中足骨底側の内側種子骨の最下点と踵骨の最下点を通り床面に垂直な断面，2）第5趾列レベル：第5中足骨頭の最下点と踵骨の最下点を通り床面に垂直な断面．各断面における上記の基準となる骨の最下点間の距離を，それぞれのレベルでのアーチ長とした．（図4）また，第1趾列レベルでは舟状骨の最上点からアーチ長の線までの最短距離をアーチ高と定義した．同様に第5趾列レベルでは立方骨の最上点からアーチ長の線までの最短距離をアーチ高とした．

縦アーチの形状（アーチの扁平程度）を評価するために，アーチ長とアーチ高の値を用いて arch height index（AHI）を算出した．これは，アーチ高をアーチ長で除して100倍したものである．内側縦アーチにおけるAHIをmedial longitudinal arch height index（MLAHI），外側縦アーチにおけるAHIを lateral longitudinal arch height index（LLAHI）とし，それぞれのレベルで荷重時，非荷重時のAHIを比較した．

統計学的解析には paired *t*-test を使用した．p値が0.05未満の場合に統計学的有意差ありとした．

結　果

可視化された足アーチ構造を荷重と非荷重条件間で比較すると，縦アーチは荷重により扁平化する方向に変化することが観察された．そしてこの変化は，主に舟状骨よりも遠位で生じ，内側でより大きい傾向にあった．（図2）縦アーチ形状の定量的評価では，MLAHIは非荷重時45.4±1.1%，荷重時40.3±0.94%で，非荷重時に比べ荷重時において有意に低下した．一方，LLAHIは非荷重時30.1±1.3%，荷重時28.6±0.93%であり，荷重時と非荷重時の間に統計学的有意差を認めなかった．

考　察

足部のアーチ構造に関する研究のうち，内側縦アーチの評価方法に関しては多くの報告がある．フットプリントはその簡便さと再現性のため，内側縦アーチを評価するのに広く用いられている．Staheli ら[3]はフットプリントのアーチ部の幅と踵部の幅との比を arch index と定義し，正常足の年齢による変化を評価した．フットプリントについては他の報告者も類似した評価方法を用いているが，計測部位が報告者により様々であり統一した評価基準がない．また，フットプリントはアーチ構造を直接可視化したものではないという限界があった．

X線写真側面像も内側縦アーチを評価するために，様々な角度が定義されて広く使用されている．Saltzman ら[4]はX線写真上での測定は骨性のアーチ構造を明確に描出できることと，再現性が高いことにより，内側縦アーチを表現する gold standard であるとした．しかし，足根骨は複雑に重なりあっているため，2次元の画像であるX線写真では正確な評価が難しく，足部に扁平足などの変形を有

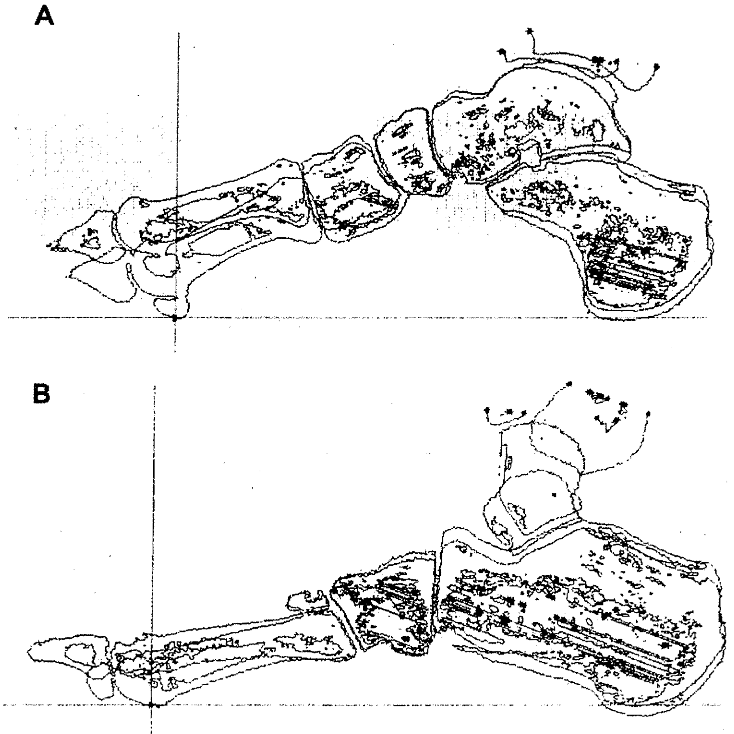

図3. 重ね合わせモデルより抽出した2つの断面
A：内側縦アーチ
B：外側縦アーチ

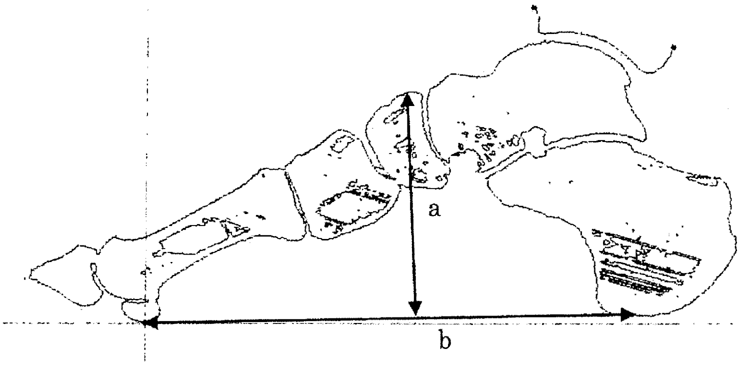

図4. アーチ高（a）とアーチ長（b）の計測部位（図は内側縦アーチの断面）

する場合にはさらに困難となる．また，外側縦アーチでは画像上骨が重なる範囲が広いため，評価法の精密さという点では疑問がある．

Nadir ら[5]は足底圧を計測し，内外側縦アーチの評価を試みた．しかしこの方法では，直接内外側アーチを可視化しておらず，さらに非荷重時とのアライメント変化をとらえることができない．

以上のように，過去に用いられてきた足縦アー

チ評価の方法には限界があり，荷重によりその構造が変化するという特徴を有する足アーチを正確に評価する方法は確立されているとはいえない．本研究では，荷重・非荷重の足部CT3次元モデルを舟状骨で重ね合わせることで，内外側縦アーチの動態を可視化した．これによって同一個体における生体足の荷重・非荷重による変化を，より正確に計測することが可能になった．

Kapandji[6]は荷重により内側縦アーチ，外側縦アーチともに扁平化すると報告した．本研究の結果でも，内外側縦アーチともに荷重により扁平化する傾向を示した．しかし，定量的評価では内側縦アーチは非荷重時に比べ荷重時に有意に扁平化したが，外側縦アーチは荷重時と非荷重時との間に統計学的有意差を認めなかった．この結果は，内側縦アーチは衝撃吸収，外側縦アーチは安定した荷重支持というように，荷重に対する主な役割が内外側で異なっていることを示唆すると考えられた．

本研究の限界として，CT撮影時の荷重の方法を荷重装置を用いた仰臥位としたため，生理的な荷重条件とは異なる可能性があり，今後の検証が必要である．

本研究で用いた方法を応用することで，内外側縦アーチに変化をきたす種々の疾患の病態解明や，より有効な治療法の開発に対して有用な情報を提供することが可能になると考えられた．

結　語

1. 非荷重・荷重条件CTデータから作製した3次元足モデルを用い，生体足の荷重による縦アーチ変化を評価した．
2. 内側縦アーチは荷重によって統計学的有意に扁平化したが，外側縦アーチは有意差はなかった．
3. 足縦アーチの荷重に対する主な役割が，内外側で異なっている可能性が示唆された．

文　献

1) Michael S, Pinzur MD. Biomechanics of the foot and ankle. Orthopaedic Knowledge Update Foot and Ankle 2008；4：3-13.
2) 木井雄一郎，鈴木智之，渡邉耕太他：足部横アーチの評価とその荷重による変化：CTによる3次元解析．北整災誌　2011；53：91-4.
3) Staheli LT, Chew DE, Corbett M. The longitudinal arch. A survey of eight hundred and eighty-two feet in normal children and adults. J Bone Joint Surg. 1987；69-A：426-8.
4) Saltzman CL, Nawoczenski DA, Talbot KD. Measurement of the medial longitudinal arch. Arch Phys Med Rehabil. 1995；76：45-9.
5) Nadir Y, Erdinc E, Ulunay K. Evaluation of the medial longitudinal arch：a comparison between the dynamic plantar pressure measurement system and radiographic analysis. Acta Orthop Traumatol Turc. 2010；44：241-45.
6) Kapandji A. L. 足底円蓋．カパンジー機能解剖学II下肢．医歯薬出版；2010. 236-61.

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論文

足関節底背屈がCTにおける距腿関節窩と脛腓靭帯結合に及ぼす影響

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要　旨　本研究の目的は足関節の肢位がCTにおける距腿関節窩と脛腓靭帯結合の離開距離に及ぼす影響を調査することである．15名の健常被験者（平均年齢29.9±8.4歳，男性11名，女性4名）において足関節最大底屈位，中間位，最大背屈位の各条件で足趾から足関節近位までCT撮影を行った．距骨滑車面から5mm遠位レベル水平面で距腿関節内側関節裂隙を計測し，脛骨天蓋レベル水平面で脛腓靭帯結合離開距離を計測した．内側関節裂隙は底屈位で平均4.9±0.8mm，中間位で平均2.7±0.5mm，背屈位で平均2.7±0.4mmで底屈位が有意に大きかった．脛腓靭帯結合離開距離は底屈位で平均2.4±0.5mm，中間位で平均3.3±0.6mm，背屈位で平均3.9±0.8mmで各足関節肢位の間で有意な差が認められた．本研究の結果より健常足関節であってもCTでは肢位によって内側関節裂隙や脛腓靭帯結合の拡大が認められた．足関節CT撮影を行う際には肢位を一定条件（中間位）に設定する必要性が示唆された．

はじめに

足関節周囲の骨折や靭帯損傷に対する診断，治療効果判定において足関節の安定性評価は重要である．三角靭帯と脛腓靭帯結合損傷の際，圧痛点や外旋ストレステストなどの理学所見とあわせてX線画像評価が必要である．この評価は撮影時の回旋肢位に大きく影響を受けるため，脛骨と腓骨が重なって撮影された場合，脛腓間離解の評価は困難である．X線像では内側関節裂隙も評価される．内側関節裂隙が5mm以上開大している場合，または距腿関節裂隙を超える場合に三角靭帯損傷と診断される[2)4)8)]．ストレスX線撮影では足関節に外旋を負荷して内側関節裂隙を評価するが，疼痛を伴うことや，外旋トルクを一定化できないことが問題である．一方，CTは骨折の診断に優れるほか，距腿関節窩と脛腓靭帯結合部の拡大を水平面で評価することができる．また足部の回旋に影響されず，再現性にも優れるという利点を有するため，足関節不安定性の評価に有用である．しかし，撮影肢位，特に足関節底背屈によるCT画像への影響は不明である．本研究の目的は，足関節底背屈がCTにおける距腿関節窩と脛腓靭帯結合部に及ぼす影響を調査することである．

対象と方法

15名の健常被検者を対象とした．男性11名，女性4名，平均年齢29.9±8.4歳（22～53歳）．平均BMIは22.4±2.6（18.8～26.8）であった．評価した15足のうち，右足が9例，左足が6例であった．

被検者には無作為に足関節中間位，最大底屈位，最大背屈位の肢位をとってもらった．各条件で足趾から下腿中間までCT撮影を行った．CTは脛骨軸に垂直に0.5mmスライスで撮影した（Aquilion 16，東芝メディカルシステムズ）．15足における3種類の足関節肢位で得られた画像について，画像解析ツールを用いて3人の整形外科医が計測を行っ

Key words：ankle mortise（距腿関節窩），syndesmosis injury（脛腓靭帯結合損傷），CT（コンピュータ断層撮影）
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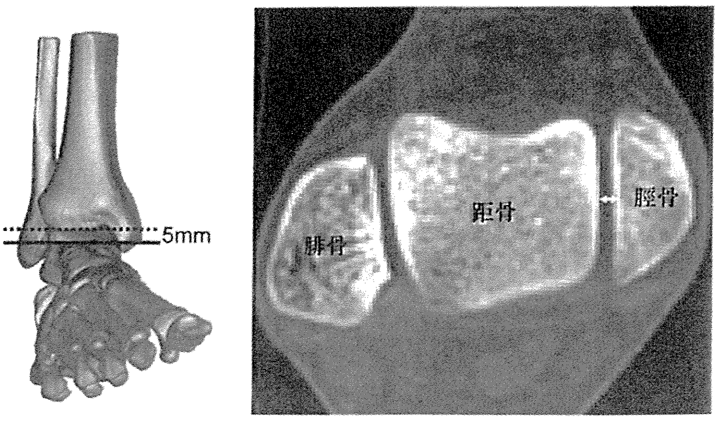

図 1 距腿関節窩（内側関節裂隙）の計測．距骨滑車面から 5 mm 遠位レベル水平面で行った．

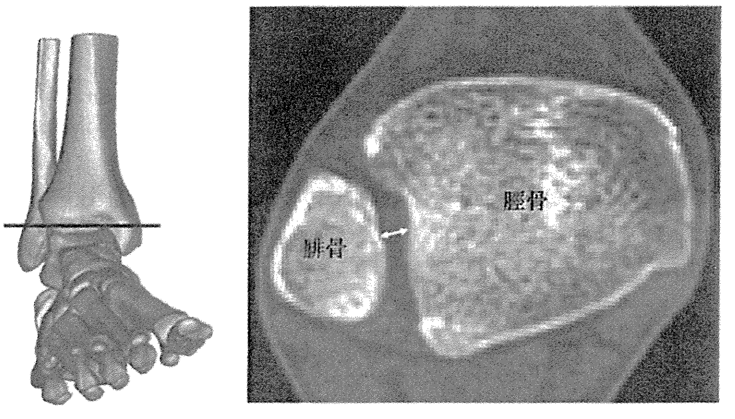

図 2 脛腓靭帯結合（脛腓間距離）の計測．脛骨天蓋レベル水平面で行った．

た．計測したのは，距骨滑車面から 5 mm 遠位レベル水平面で距腿関節の内側関節裂隙と，脛骨天蓋レベル水平面で脛腓靭帯結合離開距離である．内側関節裂隙は距骨中央部で（図 1），脛腓靭帯結合離解距離は脛骨中央部で計測した（図 2）．

3 人の計測値は平均化され，各足関節肢位の条件間で比較検討された．One-way repeated measures analysis of variance（ANOVA）で分散分析の後，Tukey test で多重比較検定された．p 値が 0.05 未満で有意差ありとした．

結　　果

CT 計測上，足関節底屈位での底屈角度は平均 44.9 ± 5.9 度，中間位では平均 2.7（底屈）± 5.1 度，背屈位での背屈角度は平均 18.8 ± 6.8 度であった．

内側関節裂隙は底屈位で平均 4.9 ± 0.8 mm，中間位で平均 2.7 ± 0.5 mm，背屈位で平均 2.7 ± 0.4 mm で底屈位が中間位と背屈位に対して有意に大きかった（それぞれ $p < 0.001$）（図 3）．

脛腓靭帯結合離開距離は底屈位で平均 2.4 ± 0.5 mm，中間位で平均 3.3 ± 0.6 mm，背屈位で平均 3.9 ± 0.8 mm で各足関節肢位の間で有意な差が認められた（底屈位 vs 中間位 $p = 0.003$，中間位 vs 背屈位 $p = 0.028$，底屈位 vs 背屈位 $p < 0.001$）（図 4）．

考　　察

脛腓靭帯結合と三角靭帯との合併損傷の頻度は，スポーツ傷害を中心に増加している[3)5)12)]．重度損傷で足関節不安定性を認める際には手術治療も選択される[1)]．遠位脛腓靭帯単独損傷であっても足関節不安定性が生じ，疼痛が残存する例も少なくない[5)10)]．診断には理学所見と画像評価が重要であり，距腿関節窩と脛腓靭帯結合の拡大が X 線写真や CT，MRI で確認されることが多い．本研究の結果では健常足

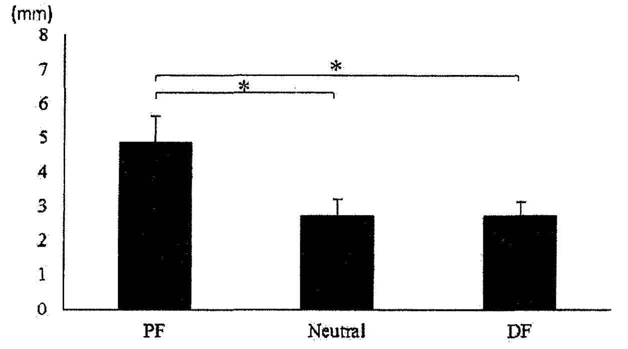

図 3 内側関節裂隙の変化．PF：底屈位，DF：背屈位．$^*p<0.05$.

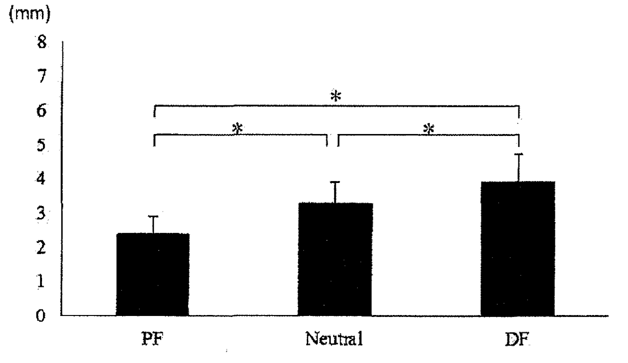

図 4 脛腓靭帯結合（脛腓間距離）の変化．PF：底屈位，DF：背屈位．$^*p<0.05$.

関節であってもCTでは肢位によって距腿関節内側関節裂隙や脛腓靭帯結合の拡大が認められた．

Saldua ら[9] は25人の足関節外傷の既往がない健常者に足関節中間位，底屈15度，30度，45度の4種類の下腿装具を使用してX線撮影を行った．その結果，底屈位になるほど内側関節裂隙は拡大し，45度底屈位では中間位と比較して0.38 mm拡大していた．彼らは保存治療が可能な症例でも底屈位撮影によって拡大が大きく認められれば手術の適応があると判定されることを危惧し，X線像は中間位で撮影することを推奨している．

今回の研究結果もSalduaらの報告と同様の傾向を示したが，こうした結果は距骨の解剖学的形状が影響していると考えられた．距骨は前方が後方よりも幅広いため[6][11]，足関節底屈位では内果関節面，外果関節面間における距骨の幅は減少し，距腿関節窩の横幅が拡大しなくても内側関節裂隙は増加するという現象が生じる．Inman[7] は距骨の幅は前方と後方で平均2.4 mmの差があることを報告しており，底屈位で内側関節裂隙が平均2.2 mm増加したわれわれの結果と近似していた．一方，背屈位では内果関節面，外果関節面間における距骨の幅は増加する．しかし，腓骨の外旋といった生理的運動も同時に生じるため，背屈するに従って距腿関節窩の横幅の拡大が生じても内側関節裂隙は中間位と変わらずに，脛腓間離開距離のみが増加したと考えられた．内側関節裂隙が中間位と背屈位との間で変化しなかったのは，中間位で既に距骨と内果関節面とが計測部位において接しているためと考えられた．

通常の足関節CT撮影時には仰臥位で自然な肢位を取るため，底屈位になっていることが多い．しかし，本研究結果のように撮影条件によって，内側関節裂隙が増加したり，脛腓間距離が変化したりするため，CT撮影は一定の条件，特に中間位で撮影する必要性が示された．

文　　献

1) Beumer, A., Valstar, E.R., Garling, E.H., et al.: Effects of ligament sectioning on the kinematics of the distal tibiofibular syndesmosis. Acta Orthop. Scand., 77 : 531-540, 2006.

2) Beumer, A., van Hemert, W.L., Niesing, R., et al. : Radiographic measurement of the distal tibiofibular syndesmosis has limited use. Clin. Orthop. Relat. Res., 423 : 227-234, 2004.
3) Boytim, M.J., Fischer, D.A., Neumann, L. : Syndesmotic ankle sprains. Am. J. Sports Med., 19 : 294-298, 1991.
4) DeAngelis, J.P., Anderson, R., DeAngelis, N.A. : Understanding the superior clear space in the adult ankle. Foot Ankle Int., 28 : 490-493, 2007.
5) Gerber, J.P., Williams, G.N., Scoville, C.R., et al. : Persistent disability associated with ankle sprains: a prospective examination of an athletic population. Foot Ankle Int., 19 : 653-660, 1998.
6) Gourineni, P.V., Knuth, A.E., Nuber, G.F. : Radiographic evaluation of the position of implants in the medial malleolus in relation to the ankle joint space: anteroposterior compared with mortise radiographs. J. Bone Joint Surg. Am., 81 : 364-369, 1999.
7) Inman, V.T. : The Joints of the Ankle. Williams & Wilkins, Baltimore, 117, 1976.
8) Park, S.S., Kubiak, E.N., Egol, K.A., et al. : Stress radiographs after ankle fracture: the effect of ankle position and deltoid ligament status on medial clear space measurements. J. Orthop. Trauma, 20 : 11-18, 2006.
9) Saldua, N.S., Harris, J.F., LeClere, L.E., et al. : Plantar flexion influences radiographic measurements of the ankle mortise. J. Bone Joint Surg. Am., 92 : 911-915, 2010.
10) Teramoto, A., Kura, H., Uchiyama, E., et al. : Three-dimensional analysis of ankle instability after tibiofibular syndesmosis injuries: a biomechanical experimental study. Am. J. Sports. Med., 36 : 348-352, 2008.
11) Tornetta, P. 3rd, Spoo, J.E., Reynolds, F.A., et al. : Overtightening of the ankle syndesmosis: is it really possible? J. Bone Joint Surg. Am., 83 : 489-492, 2001.
12) Wright, R.W., Barile, R.J., Surprenant, D.A., et al. : Ankle syndesmosis sprains in National Hockey League players. Am. J. Sports. Med., 32 : 1941-1945, 2004.

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Effects of Dorsi and Plantar Flexion of the Ankle on the Ankle Mortise and Syndesmosis in CT Imaging

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Abstract

The purpose of this study was to investigate the effects of dorsi and plantar flexion of the ankle on the ankle mortise and syndesmosis in CT imaging.

CT images were taken of fifteen healthy volunteers with the ankle in three positions: plantar flexion, neutral position, and dorsi flexion. Three observers measured ankle medial joint space and syndesmosis diastasis in each CT image.

The mean medial joint space was 4.9 ± 0.8 mm with ankle plantar flexion, 2.7 ± 0.5 mm in neutral position, and 2.7 ± 0.4 mm with dorsi flexion. Ankle plantar flexion significantly increased the medial joint space (p<0.001). Syndesmosis diastasis was 2.4 ± 0.5 mm with ankle plantar flexion, 3.3 ± 0.6 mm in neutral position, and 3.9 ± 0.8 mm with dorsi flexion. Significant differences were observed between plantar flexion and neutral position (p=0.003), neutral and dorsi flexion (p=0.028), and plantar and dorsi flexion (p<0.001).

Dorsi and plantar flexion of even the normal ankle changed the ankle medial joint space and syndesmosis diastasis in CT images. Thus, the ankle should be maintained in a neutral position for diagnosis of deltoid ligament and syndesmosis injuries in CT imaging.

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ORIGINAL PAPER

Analysis of ankle–hindfoot stability in patients with ankle instability and normals

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Abstract
Purpose We devised a testing apparatus for in vivo analysis of ankle stability. The purpose of the study was to test the reliability of this apparatus and to determine the stability pattern of the ankle–hindfoot complex in healthy, asymptomatic volunteers and in patients with ankle instability.
Methods Ten healthy individuals were studied, and testing was repeated on the same day and different days. Three patients with symptomatic, unstable ankles were also tested on both involved and uninvolved sides. Constant inversion torque was applied, then internal rotation torque, while moving the ankle throughout the range of sagittal motion. Three-dimensional kinematics of the ankle–hindfoot complex were measured by an electromagnetic tracking system.
Results Measurements were repeatable, with intraclass correlation coefficients 0.9 or better. Variability was observed among controls, but motion curve patterns were consistent. Motion curve slopes were sensitive in differentiating between unstable and stable ankles.
Conclusions Most previous reports are in vitro studies conducted with the ankle in one position, manual stress applied, or joint positions estimated with planar radiographs. Our study indicated that more accurate diagnosis of severity of ankle ligament injuries may be possible.

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Introduction

Ankle ligament sprains are common injuries. One epidemiological study revealed that the estimated incidence rate of ankle sprains in the general population presenting to emergency departments in the United States is 2.15 per 1,000 person-years, and nearly half of all ankle sprains occurred during athletic activity [1]. The most common risk factor for ankle sprains in sports is a previous history of an ankle sprain [2]. Yeung et al [3] reported the recurrence rate of ankle sprains for athletes with a previous history of ankle sprains was as high as 73%. Progression to chronic problems, such as pain, giving-way, instability, and ultimately degenerative changes is not uncommon in patients with previous ankle sprains [3, 4].

A clear understanding of ankle–hindfoot motion is important for both evaluation and treatment of ankle disorders. Many investigators have studied and reported the roles of ankle structures in joint stability and mobility, which are functions of both extrinsic (i.e., ligaments) and intrinsic (i.e., articular geometry) elements. Studies have been conducted to determine the range of motion, contribution of lateral ligaments to stability, and the effects of ligament rupture on mobility in both in vitro and in vivo experiments. In spite of these efforts, the clinical assessment of patients with ankle instability still has some inaccuracies [5, 6]. Evaluation of ankle–hindfoot stability under applied force, such as inversion or anterior translation, is widely performed clinically and radiologically. It is known that these evaluating methods have some limitations. Stress radiographs of the ankle measure two-dimensional displacement, but instability occurs in three planes. Moreover, in these stress evaluations and earlier studies, ankle-joint stability was investigated in discrete joint positions. However, the positions selected may or may not have been positions in which laxity was at its maximum. The optimal joint position for testing varies depending upon the specific ligament being tested. Also, most of these investigations were cadaveric studies that used methods not directly applicable to testing patients.

We developed an ankle-testing device to measure three-dimensional ankle and hindfoot motion with a specific rotational force applied and succeeded in distinguishing between controls and injured ankles and isolated the anterior talofibular ligament (ATFL) and combined ATFL/calcaneofibular ligament (CFL) laxity in vitro [7–9]. The purposes of this study were to: (1) determine the repeatability of the testing methods in vivo; (2) apply this technique to uninjured controls; (3) apply this technique to patients with chronic lateral ankle ligament instability to determine the feasibility of its use in routine clinical testing.

Materials and methods

Patient profile

Ten individuals without previous foot trauma or pathology were tested. Three were women; average age was 35 (range 26–42) years. The tests were performed three to five times on two different days in order to assess the repeatability of the method. Three patients with unilateral injuries to the lateral ligaments were also tested. Mean age was 26 (range 14–41) years. Two were women. All had chronic instability based upon physical examination with a positive anterior drawer, and all had abnormal stress radiographs. Both the unstable and stable ankles were tested in the device. All three patients had combined ATFL/CFL rupture confirmed at the time of surgery. Patients underwent reconstruction of lateral ankle ligament using the modified Broström procedure.

Ankle-stability-testing device

The ankle-testing apparatus was constructed primarily of acrylic plastic (Fig. 1). It allowed three rotations (internal/external, inversion/eversion, plantar/dorsiflexion) of the footplate and three translations (anterior/posterior, medial/lateral, proximal/distal) in a global anatomical coordinate system of the hindfoot. In this coordinate system, the X axis was along the tibial shaft through the centre of the ankle. The Z axis was parallel to the projection of a line connecting the centre of the heel and the second metatarsal on a plane perpendicular to the X axis. The Y axis was the product of the X axis and Z axis following the right-hand rule, passing through the ankle-joint centre. The X axis was defined as the internal/external rotation axis, the Z axis as the inversion/eversion axis, and the Y axis as the plantar/dorsiflexion axis. A 1.7 Nm torque was applied separately to each axis using a stainless steel, 180° torsion spring.

Each patient was placed in the apparatus in a seated position in 90° of hip and 90° of knee flexion. The lower leg was fixed on a low-friction sliding block to allow free

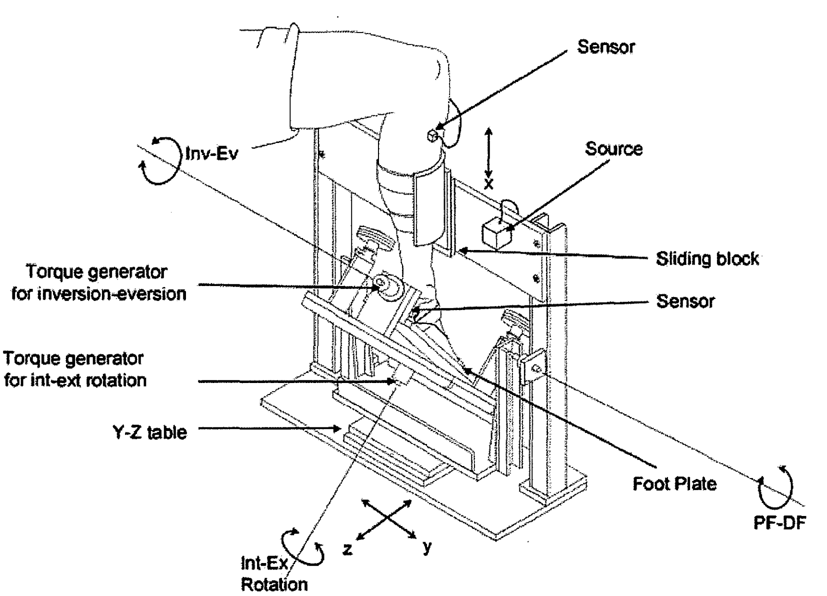

Fig. 1 Ankle-testing apparatus

proximal–distal motion of the tibia. The foot was fixed on the foot plate with a Velcro® strap on the forefoot and midfoot. The hindfoot was secured to a heel cup on the plate by a Velcro® strap. Three-dimensional movement of the calcaneus relative to the tibia (ankle–hindfoot complex) was monitored with an electromagnetic tracking system (Flock of Birds®, Ascension Technology, Burlington, VT, USA), which provided 6 degrees of freedom tracking. The magnetic source was mounted to the frame of the testing device. One sensor was attached to the skin overlying the calcaneal tuberosity, and another was attached to the anteromedial aspect of the proximal leg. Placement of sensors in these areas with less subcutaneous tissues limited the potential for artifact from skin movement. A pilot study was performed to compare kinematics data from both skin- and bone-mounted sensors using cadaver feet, which confirmed that the measurements from skin-mounted sensors correlated well with bony movements [7].

Internal rotation then inversion stresses were applied using constant loading of the foot with a built-in, spring-loaded torque generator. A constant torque of 1.7 Nm was applied, determined on the basis of previous tests and clinical experience. The foot was moved slowly from maximum plantarflexion to maximum dorsiflexion, with the constant torque applied to the foot in either internal rotation or inversion. Three-dimensional kinematic data were recorded continuously during testing. Internal/external rotation and inversion/eversion movement curves expressed as functions of the plantar/dorsiflexion angle. Internal rotation and inversion angles and the slope of the curves were determined for the unstable and the opposite stable ankle, as well as for controls. Discrete points were selected for data analysis: the angle of each curve was determined at 20° and 10° of plantarflexion, neutral position, and 10° of dorsiflexion. The slope was calculated between the points representing maximum motion and a point corresponding to motion at 10° of dorsiflexion from the internal rotation curve or the inversion curve.

Statistical analysis was performed using a paired t test to evaluate the difference between stable and unstable ankles, with statistical significance set at $p<0.05$. An unpaired t test was used to evaluate the difference between unstable ankles and controls, with statistical significance set at $p<0.05$ level.

Results

Repeatability

The technique had repeatable results. The intraclass correlation (ICC) > was 0.97 for same-day testing and 0.95 for testing on different days for the inversion stress test. For the internal rotation stress test, the ICC was 0.93 for intraday tests and 0.90 for interday tests. An example of three trials obtained from one individual is shown in Fig. 2.

Uninjured control ankles

Angular data obtained from ten controls are shown in Fig. 3. Maximum inversion was observed in a position >20° of plantarflexion and decreased as the ankle dorsiflexed. The mean ± standard deviation (SD) value of maximum inversion was 15.8°±3.7° , with a range of 8.3–22.5°. Mean value of minimum inversion was 6.6°±2.6°, with a range of 2.2–10.6°. Mean maximum to minimum inversion was 9.2°± 3.2°, with a range of 2.7–13.4°. The shape of curves for the internal rotation stress test was convex with the peak of the convex curve at an average of 7.1°±6.1° plantarflexion, ranging from 17° plantarflexion to 1° dorsiflexion. Mean value of maximum internal rotation was 15.4°±3.4°, with a range of 8.8–20.7°. Mean value of minimum internal rotation was 11.7°±3.5°, with a range of 6.5–18.2°. Mean range from maximum to minimum internal rotation was 3.7°±1.9°, with a range of 1.2–7.4°.

Unstable ankles

The motion curve patterns in unstable ankles were similar among the three patients with lateral ankle instability.

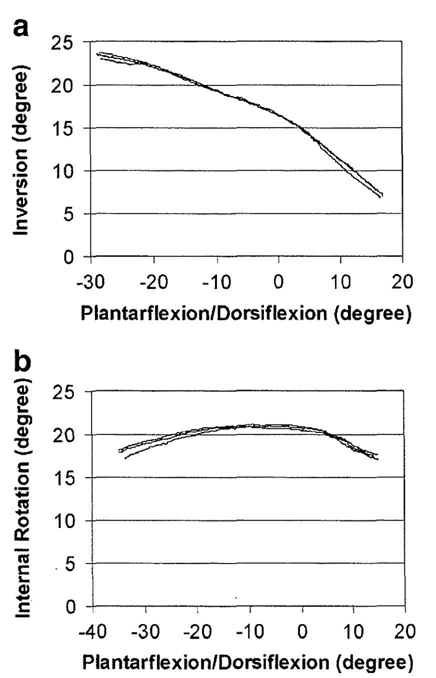

Fig. 2 Ankle–hindfoot complex motion curve of an uninjured control anke: three trials. Inversion or internal rotation motion of the ankle–hindfoot complex is expressed as a function of plantar/dorsiflexion angle. **a** Inversion stress test. **b** Internal rotation stress test

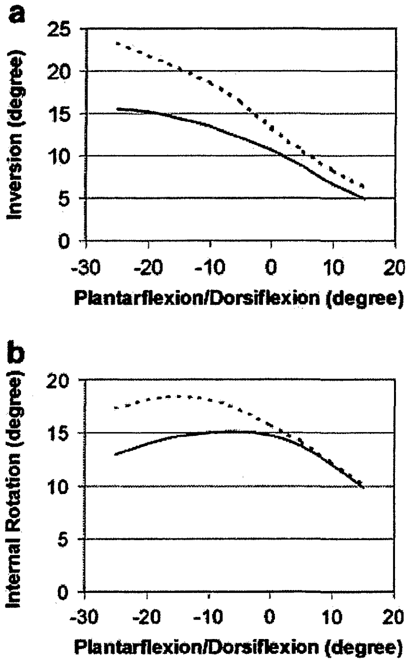

Fig. 3 Mean ankle–hindfoot complex motion showing the effect of ankle-flexion angle on the amount of inversion or internal rotation in ten uninjured controls (*solid lines*) and three patients with instability (*dotted lines*). **a** Inversion stress test. **b** Internal rotation stress test

Overall, there was more motion with inversion and internal rotation testing. The degree of rotation at five discrete points was identified and compared between the unstable and the stable ankle. These points were 20° and 10° of plantarflexion, neutral, 10° of dorsiflexion, and maximum internal rotation or inversion position. For the inversion stress test, the difference in motion curves was greatest in plantarflexion. Further analysis of the motion curve slopes was much more sensitive in differentiating between the unstable and the opposite stable ankle (Figs. 4 and 5). With internal rotation force applied, mean ± SD slope was 29.4±8.3

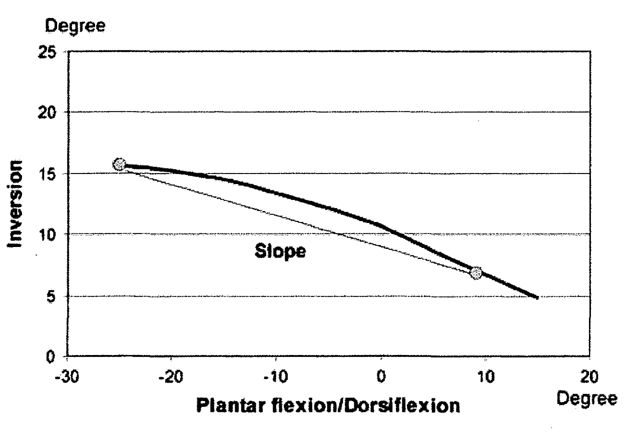

Fig. 4 Determination of slope of the motion curve from a point at 10° of dorsiflexion to a point at the peak of the motion curve

(in the unstable ankle, which was significantly greater than 11.7±5.0 in the opposite stable ankle (p=0.03). With inversion force applied, the mean ± SD slope of 44.8±11.4 in the unstable ankle was significantly different than that of 20.7±2.4 in the opposite stable ankle (p=0.04). Similarly, slopes of the inversion and internal rotation motion curves were significantly greater in unstable ankles than in uninjured controls (Fig. 5).

Discussion

Stress radiography with anterior drawer or inversion stress is widely accepted as a means to assess patients with lateral ankle ligament injuries. Anterior talar displacement or talar tilt angle is measured under stress loading. However, accuracy of these conventional stress tests is suspect [4–6, 10–15]. Some limitations in the standard evaluation methods were because ankles were tested in various

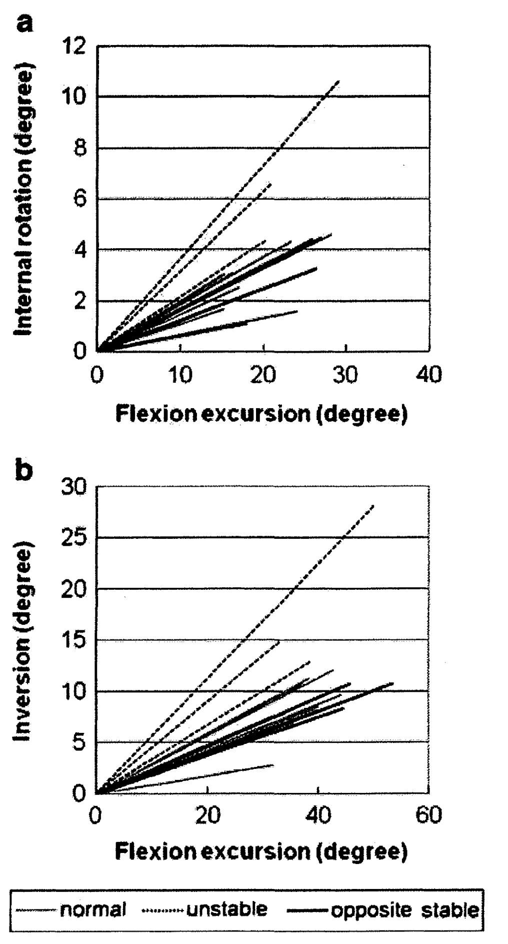

Fig. 5 Slopes from motion curves in three unstable ankles (*dotted lines*), their opposite stable ankles (*solid black lines*), and ten uninjured controls (*solid gray lines*). **a** Internal rotation stress test. **b** Inversion stress test

positions and with different forces applied. Different authors recommended stressing the ankle in dorsiflexion, in the neutral position, or in plantarflexion during the stress manoeuvre [6, 10, 16, 17]. The foot position affects ankle stability because the relationship between ligamentous laxity and bony constraints vary with flexion angle during stress testing [7, 9, 18–21]. Our results showed that mobility of the ankle–hindfoot complex varied as foot flexion angle changed. The magnitude of inversion increased with increasing plantarflexion position of the ankle under inversion stress.

The diagnostic method in this study is based upon foot-motion measurements under different specific manipulations and analysis of motion curve patterns using trend analysis [9]. This can be a fundamental improvement over the existing stress testing, which depends upon a discrete one-point value to determine laxity and type of injury. Each ligament contributes to joint stability at a certain joint position, and the rupture of a ligament will cause instability at that position. Thus, comparison based on the overall curve trend should be more sensitive than using discrete points in evaluating ligament injury. The existing testing analyses rotation in one plane (inversion) or translation in one direction (anterior drawer) but does not consider other planes of rotation or displacement, such as internal rotation. It may be reasonable to measure internal rotation to diagnose ATFL injury because the ATFL functions primarily in restricting internal rotation of the talus in the mortise [22]. Although this device may be applicable to stress radiography, we think that the motion-curve analysis during full range of sagittal motion will provide more detailed information of ankle–hindfoot complex kinematics and lead to more accurate diagnosis of severity of ankle ligament injuries.

We calculated slopes from the motion curves and applied them to differentiate between unstable (combined ATFL and CFL rupture) and stable ankles because a previous in vitro study demonstrated that it was possible to differentiate simulated ATFL injury from an intact ankle and ATFL/CFL injury from ATFL injury with this method [9]. Further study is needed of patients who have persistent ankle pain after sprains in order to determine whether these distinctions can be made in vivo; such analysis is not possible with conventional stress tests. Also, there is a subset of patients with persistent pain after ankle ligament reconstruction or after injury who have negative stress radiographs and stable ankles on examination who are considered to have "functional instability." Perhaps the diagnoses of some of these patients can be clarified with the testing method described here. We tested patients with inversion or internal rotation force applied. However, with the testing device, it is also possible to apply constant eversion, external rotation, combined inversion/internal rotation, and eversion/external rotation stress to the ankles. It may therefore be an applicable assessment for patients with other diagnoses, such as suspected deltoid ligament instability.

Although this is a promising noninvasive method of differentiating ankle ligament instabilities, there are limitations. We found variability in absolute values of inversion or internal rotation measured between patients. It is well recognised that there is wide variation among individuals in stress tests [6, 12, 13, 23], which our testing confirmed. This observation in preliminary testing led to refinements in how the foot was secured to the footplate, standardising a neutral position prior to testing, and the repeatability study that was conducted. Data analysis was time consuming, and a special device to measure continuous three-dimensional movement is needed. Some modifications will be required when this device is considered as a widely used examination tool in clinical settings.

Patients with unstable ankles were differentiated from uninjured controls by using the ankle-stability-testing device. The method is noninvasive, does not involve radiation exposure, and is repeatable. It has the potential of providing more accurate diagnosis of ankle-ligament injuries.

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Conflict of interest The authors declare that they have no conflict of interest.

References

1. Waterman BR, Owens BD, Davey S, Zacchilli MA, Belmont PJ Jr (2010) The epidemiology of ankle sprains in the United States. J Bone Joint Surg Am 92:2279–2284. doi:10.2106/JBJS.I.01537
2. Thacker SB, Stroup DF, Branche CM, Gilchrist J, Goodman RA, Weitman EA (1999) The prevention of ankle sprains in sports. A systematic review of the literature. Am J Sports Med 27:753–760
3. Yeung MS, Chan KM, So CH, Yuan WY (1994) An epidemiological survey on ankle sprain. Br J Sports Med 28:112–116
4. Karlsson J, Lansinger O (1992) Lateral instability of the ankle joint. Clin Orthop Relat Res:253–261
5. Frost SC, Amendola A (1999) Is stress radiography necessary in the diagnosis of acute or chronic ankle instability? Clin J Sport Med 9:40–45
6. Fujii T, Luo ZP, Kitaoka HB, An KN (2000) The manual stress test may not be sufficient to differentiate ankle ligament injuries. Clin Biomech (Bristol, Avon) 15:619–623. doi:S0268-0033(00)00020-6
7. Fujii T, Kitaoka HB, Luo ZP, Kura H, An KN (2005) Analysis of ankle-hindfoot stability in multiple planes: an in vitro study. Foot Ankle Int 26:633–637. doi:894987
8. Fujii T, Kitaoka HB, Watanabe K, Luo ZP, An KN (2006) Comparison of modified Brostrom and Evans procedures in simulated lateral ankle injury. Med Sci Sports Exerc 38:1025–1031. doi:10.1249/01.mss.0000222827.56982.40

9. Fujii T, Kitaoka HB, Watanabe K, Luo ZP, An KN (2010) Ankle stability in simulated lateral ankle ligament injuries. Foot Ankle Int 31:531–537. doi:10.3113/FAI.2010.0531
10. Bahr R, Pena F, Shine J, Lew WD, Lindquist C, Tyrdal S, Engebretsen L (1997) Mechanics of the anterior drawer and talar tilt tests. A cadaveric study of lateral ligament injuries of the ankle. Acta Orthop Scand 68:435–441
11. Becker HP, Komischke A, Danz B, Bensel R, Claes L (1993) Stress diagnostics of the sprained ankle: evaluation of the anterior drawer test with and without anesthesia. Foot Ankle 14:459–464
12. Cass JR, Morrey BF, Chao EY (1984) Three-dimensional kinematics of ankle instability following serial sectioning of lateral collateral ligaments. Foot Ankle 5:142–149
13. Glasgow M, Jackson A, Jamieson AM (1980) Instability of the ankle after injury to the lateral ligament. J Bone Joint Surg Br 62-B:196–200
14. Johannsen A (1978) Radiological diagnosis of lateral ligament lesion of the ankle. A comparison between talar tilt and anterior drawer sign. Acta Orthop Scand 49:295–301
15. Larsen E (1986) Experimental instability of the ankle. A radiographic investigation. Clin Orthop Relat Res:193–200
16. Grace DL (1984) Lateral ankle ligament injuries. Inversion and anterior stress radiography. Clin Orthop Relat Res:153–159
17. Lahde S, Putkonen M, Puranen J, Raatikainen T (1988) Examination of the sprained ankle: anterior drawer test or arthrography? Eur J Radiol 8:255–257
18. Allinger TL, Engsberg JR (1993) A method to determine the range of motion of the ankle joint complex, in vivo. J Biomech 26:69–76. doi:0021-9290(93)90614-K
19. Blanshard KS, Finlay DB, Scott DJ, Ley CC, Siggins D, Allen MJ (1986) A radiological analysis of lateral ligament injuries of the ankle. Clin Radiol 37:247–251. doi:S0009-9260(86)80328-2
20. Hollis JM, Blasier RD, Flahiff CM (1995) Simulated lateral ankle ligamentous injury. Change in ankle stability. Am J Sports Med 23:672–677
21. Kovaleski JE, Gurchiek LR, Heitman RJ, Hollis JM, Pearsall AWT (1999) Instrumented measurement of anteroposterior and inversion-eversion laxity of the normal ankle joint complex. Foot Ankle Int 20:808–814
22. Rasmussen O (1985) Stability of the ankle joint. Analysis of the function and traumatology of the ankle ligaments. Acta Orthop Scand Suppl 211:1–75
23. Lapointe SJ, Siegler S, Hillstrom H, Nobilini RR, Mlodzienski A, Techner L (1997) Changes in the flexibility characteristics of the ankle complex due to damage to the lateral collateral ligaments: an in vitro and in vivo study. J Orthop Res 15:331–341. doi:10.1002/jor.1100150304

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The role of ankle ligaments and articular geometry in stabilizing the ankle

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ABSTRACT

Background: Ankle joint stability is a function of multiple factors, but it is unclear to what extent extrinsic factors such as ligaments and intrinsic elements such as geometry of the articular surfaces play a role. The purposes of this study were to determine the contribution of the ligaments and the articular geometry to ankle stability and to determine the effects of ankle position and simulated physiological loading upon ankle stability.
Methods: Sixteen cadaveric lower extremities were studied in unloaded and with axial load equivalent to body weight. Anterior–posterior, medial–lateral translation and internal–external rotation tests were performed in neutral, dorsiflexion and plantarflexion ankle positions. Intact ankle stability was measured; ankle ligaments were serially sectioned and retested.
Findings: For unloaded condition, the lateral ligament accounted for 70% to 80% of anterior stability and the deltoid ligaments for 50% to 80% of posterior stability. Both ligaments contributed 50% to 80% to rotational stability; however, the ligaments did not provide the primary restraints to medial–lateral stability. For loaded ankle condition, articular geometry contributed 100% to translational and 60% to rotational stability. The ankle was less stable in plantarflexion and more stable in dorsiflexion.
Interpretation: The contribution of extrinsic and intrinsic elements to ankle stability is dependent upon the load and direction of force applied. This study underscores the importance of restoring soft tissues about the ankle to the anatomic condition during reconstruction operations for instability, trauma and arthritis.

1. Introduction

The ankle joint is affected by a variety of disorders such as sprains, fractures and arthritis (Mack, 1982; Yeung et al., 1994). The estimated incidence rate of ankle sprains in the general population presenting to emergency departments in the United States is 2.15 per 1000 person-years (Waterman et al., 2010). In the sports field, ankle sprains account for 20% to 25% of all time-loss injuries in running or jumping sports such as basketball, football, soccer, field hockey and volleyball (Mack, 1982). Twisting of the ankle under load may cause ankle fractures, ligamentous injuries, or both. Specific injuries depend on several factors such as weightbearing, orientation of the forces involved, or stabilizing roles of ankle structures.

Biomechanical studies have been performed on the ankle joint, specifically focusing on ankle stability. Ankle stability is a function of both extrinsic elements such as ligaments and intrinsic elements such as geometry of the articular surface. The contribution of extrinsic factors has been studied in joints (Butler et al., 1980; Itoi et al., 2000; Knutson et al., 2000; Minami et al., 1985; Morrey and An, 1983; Ritt et al., 1998) including the ankle (Stormont et al., 1985; Tochigi et al., 2006). A previous study reported contribution of the ligaments and articular surface in the ankle to rotation and version stability, but did not include translational stability (Stormont et al., 1985). Another study reported contribution of the articular surface geometry to ankle stability, but did not include ligamentous contribution (Tochigi et al., 2006). There is still uncertainty about the effects of weightbearing, different ankle positions and soft tissue integrity upon ankle stability, as well as the contribution of intrinsic factors such as joint geometry. With lower levels of loading of the ankle, it is presumed that ligamentous structures function as more dominant stabilizers. As axial load increases, the contribution of the articular geometry to stability is presumed to increase. These relationships are relevant to understanding the mechanisms of joint injury, as well as surgical interventions such as ligament reconstruction and arthroplasty.

Methods for measuring intrinsic joint stability as well as ligamentous contributions have been reported from this laboratory in various joints (Berglund et al., 1994; Halder et al., 2001; Haugstvedt et al., 2002; Itoi et al., 2000; Stuart et al., 2000), and established in the ankle joint (Watanabe et al., 2009). The purpose of this study was to determine the contribution of the deltoid and lateral ankle ligaments as well as articular geometry to stability of the ankle joint and to determine the effects of ankle position and simulated physiological loading upon ankle stability.

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2. Methods

There were two parts of the study. In the first part, sixteen normal (uninjured), fresh-frozen human cadaveric lower extremities donated from 13 individuals without foot–ankle pathology were studied. The mean age of the specimens was 72 years (range, 61 to 83 years). Fourteen specimens were male and two were female. Eight were right feet and eight were left feet. In the second part, eight specimens donated from 6 individuals were tested under simulated ankle ligament injury conditions. Seven specimens were male and one was female. Six were right feet and two were left feet. The specimens were disarticulated at the knee and at the mid-tarsal joints. Soft tissues including skin, subcutaneous tissues and muscles were dissected, maintaining all ligaments and the interosseous membrane intact. The subtalar joint was immobilized by Steinmann pins and hindfoot potted in polymethylmethacrylate (PMMA) cement in the neutral position. The institutional research ethics committee reviewed and approved the study.

The testing device provided simultaneous measurement of multiaxis loading and displacements of the talus relative to the tibia (Fig. 1). The device was designed in order to allow for testing in multiple planes without removing or repositioning the specimens for various directions of translation. It was designed to permit operative procedures to be performed without unmounting and remounting the specimen for each testing condition, thus reducing the potential for artifact. The tibia, but not the fibula, was potted and fixed to the testing frame to permit tibiofibular motion. A 6-component load cell (AMTI, Newton, MA, USA) was used to measure forces and moments. The hindfoot (talus, calcaneus) was potted in a rectangular metal fixture. This fixture was attached to two *X–Y* motorized horizontal stages for applying medial–lateral (ML) and anterior–posterior (AP) translation forces and a motorized rotary stage for applying internal or external rotation motions. An additional low friction *x-y* positioning slide was placed beneath the motorized stages to allow for proper positioning of the hindfoot relative to the tibia and to minimize any constraint in specific axis during the testing regimen. During AP translation testing, the AP positioning slide was locked, but ML translation was unconstrained. During ML translation testing, the ML positioning slide was locked, but AP translation was unconstrained. During rotation testing, both AP and ML positioning slides were released to allow unconstrained pure moments about that axis. Linear potentiometers (Novotechnik, Southborough, MA) were used to measure translational displacements at the joint level and were affixed between the potted hindfoot and distal tibia. Internal–external rotation (adduction, abduction) of the ankle was measured with a rotatory potentiometer incorporated in the motorized rotary stage. Internal–external rotation occurred about the axis along the tibial shaft which was perpendicular to the *X–Y* horizontal stage.

Motor control and data acquisition were accomplished with Labview software (National Instruments, Austin, TX, USA). The frame to which the tibia was attached was clamped onto a low-friction vertical slide. The tibia was suspended to this frame with multiple heavy threaded Steinman pins, but the fibula was not fixed, in order to permit physiological tibio-fibular motion. Two axial loading conditions were tested: a minimal level (5 N) and approximately body weight (700 N) through this slide by a low-friction pneumatic cylinder (Illinois Pneumatic, Roscoe, IL, USA). The ankle bears high levels of compressive forces during normal level walking (Simonsen et al., 1995), but in preliminary experiments, we found that repeated testing of anatomic specimens at high loading levels could alter the soft tissue and bony anatomy. In a pilot study, ankles were tested in 100 N axial load increments up to 1000 N and we found that the load-displacement characteristics of the talus relative to the tibia became consistent at the 600 N level and higher. Therefore, 700 N was chosen as the simulated physiological axial loading of body weight. Axial loads of 5 N (essentially unloaded) and 700 N (loaded) were applied in each of three ankle positions: 15° of plantarflexion, neutral and 10° of dorsiflexion. These angles represented the range of sagittal ankle motion during the stance phase of gait. Experimentally, these positions were consistently obtained by inserting a wedge (10° or 15°) under the ankle fixture.

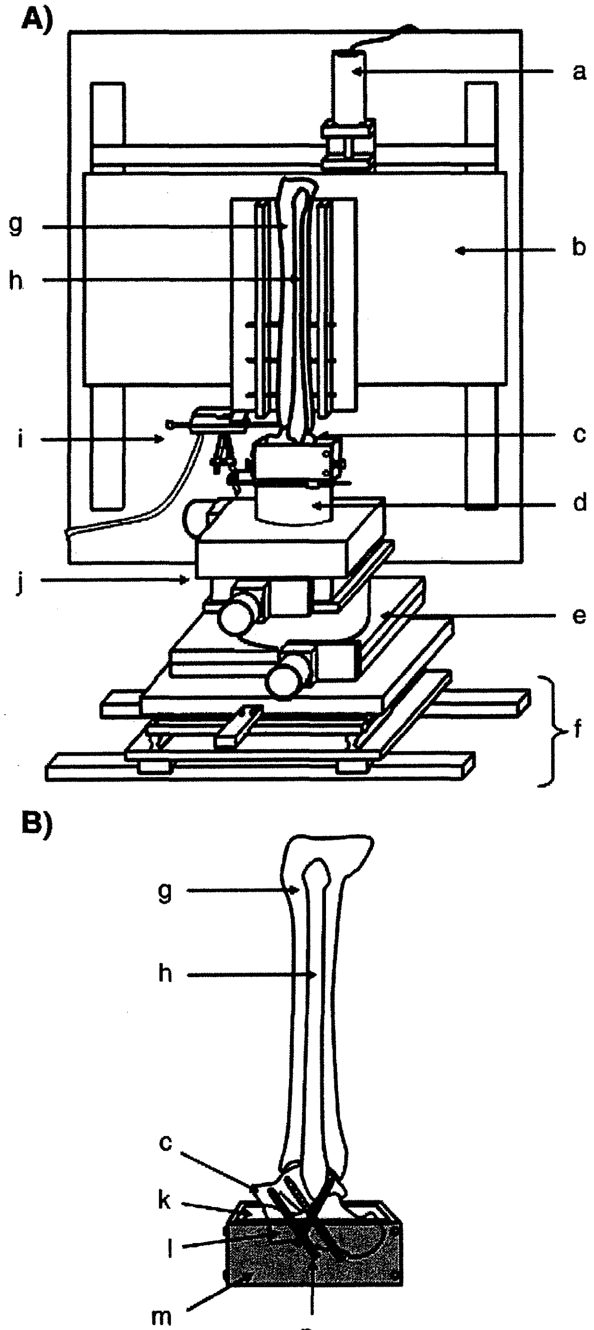

Fig. 1. Multi-axis materials testing machine: A, testing device; B, lower extremity specimen embedded in PMMA. a) Axial load actuator, b) low-friction vertical slide, c) talus, d) six component load cell, e) motorized rotary stage, f) positioning stage, g) tibia, h) fibula, i) potentiometer, j) motorized horizontal stage, k) polymethylmethacrylate, l) calcaneus, m) potting fixture (gray), n) Steinmann pins. The tibial shaft was fixed perpendicular to the *X–Y* horizontal stage (j) and internal–external rotation was about the axis along the tibial shaft.

All anatomic specimens were tested in the normal, intact condition for the first part of the study. In the second part, further testing was performed in eight feet. In four feet, the deltoid ligament was sectioned first and testing repeated, then the lateral ankle ligaments were sectioned and testing performed. In another four feet, the lateral ankle ligaments were sectioned and testing performed, then the deltoid ligament and testing completed. The tibiofibular syndesmosis, anterior tibiofibular and posterior tibiofibular ligaments proximally and distally were left intact in order to permit free movement between the tibia and fibula.

Translation tests consisted of anterior, posterior, medial and lateral translation of the talus at a rate of 2 mm per second. Translation was described relative to the global horizontal rather than to the foot's local segment horizontal. When the ankle was dorsiflexed and plantarflexed, the anterior–posterior translation was determined with respect to the global horizontal or the X–Y plane of the testing apparatus. For intact specimens, after a load limit of 150 N in the direction of motion was reached, the direction was reversed and testing completed. For initial rotation testing of intact specimens, internal or external rotation torque was applied to the talus at a rate of 2 degrees per second up to the torque limit of 2.5 Nm. After the ligament sectioning, these tests were repeated; however translational and rotational displacement values were used as endpoint limits for each specimen. The limits for translation tests were determined as the displacement in each direction corresponded to that measured under 150 N of load in the intact condition. The limits of rotation corresponded to that measured at 2.5 Nm of torque in the intact specimen.

2.1. Data analysis

Testing conditions included the intact foot, sectioning the lateral ligaments or deltoid ligament and sectioning both ligaments in each of the three ankle positions (15° of plantarflexion, neutral and 10° of dorsiflexion). All tests were performed for both the 5 N (unloaded) and 700 N (loaded) axial load conditions. Load–displacement curves were obtained from the intact, one ligament sectioned and both ligaments sectioned conditions. An example is seen in Fig. 2. The corresponding forces at the predetermined displacement values were derived from each curve. Relative contributions of the sectioned ligaments to joint stability were expressed as a percentage and calculated as the reduction in load from the intact, divided by the intact load value. Contribution of articular geometry to joint stability was expressed as a percentage and calculated as the remaining load after both ligaments were sectioned, divided by the intact load value. For the purposes of this study, we defined a primary restraint structure as one that provides more than 50% of restraint to translation or rotation.

Statistical analysis was performed with repeated measures ANOVA and Friedman test. The level of significance was set at $P<0.05$. Significant main effects were further analyzed using the Student–Newman–Keuls multiple comparisons test.

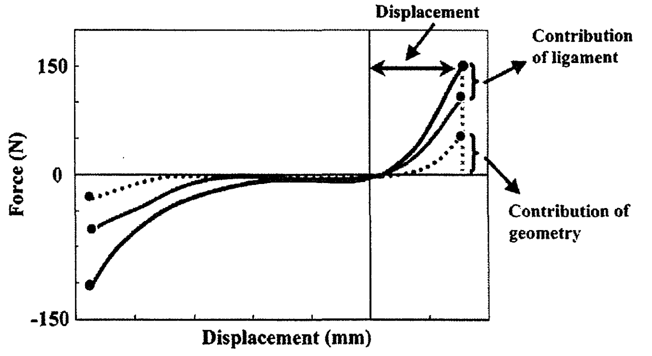

Fig. 2. Typical force–displacement curve in 5 N axial loading condition. The solid line represents the intact condition, the gray line represents one ligament sectioned and dotted line represents both ligaments sectioned. Tibiotalar displacement and contribution supporting elements are shown.

3. Results

The average and standard deviation of the absolute values measured at the talus at prescribed maximum load or torque values for all movement directions for the intact foot condition are shown in Fig. 3. Ankle position altered laxity of the ankle. The plantarflexed position was less stable and the dorsiflexed position most stable. With the unloaded ankle in plantarflexion, there was more movement in posterior, medial and lateral displacement as well as internal rotation than in dorsiflexion. In the dorsiflexion position, there was less posterior, medial and lateral displacement, as well as external and internal rotation than neutral position.

The effects of ankle position were also apparent when the ankles were loaded. With the loaded ankle in plantarflexion, there was more movement in medial and lateral displacement as well as internal and external rotation than in dorsiflexion. In the dorsiflexion position, there was less lateral displacement, as well as external and internal rotation than neutral position.

The relative contributions of the ligaments and the articular geometry to joint stability are shown in Figs. 4 and 5. In the unloaded ankle, the lateral ligaments were the primary restraint to anterior translation with the relative contribution ranging from 71% to 81%. The primary restraint to posterior translation was the deltoid ligament, with relative contribution ranging from 50% to 80%. The contribution of the deltoid ligament was significantly higher in plantarflexion than in dorsiflexion ($P<0.05$). For medial–lateral translation, neither of the two ligaments served as a primary restraint. For internal–external rotation, both the lateral and the deltoid ligaments contributed in stabilizing the ankle, The contribution of the deltoid ligament to external rotation was significantly lower in dorsiflexion than in neutral and plantarflexion ($P<0.05$).

The effect of ankle loading and articular geometry was apparent with testing after sectioning one or both ankle ligaments. The loaded ankle remained stable after ligament sectioning in anterior–posterior and medial–lateral translation with little change in the load displacement curves. Similarly, the loaded ankle remained stable with rotation. The articular geometry was the primary restraint with relative contributions ranging 50% to 74% (Fig. 5). We were unable to detect a difference in relative contribution between the three ankle positions.

4. Discussion

The contribution of extrinsic and intrinsic elements to ankle stability was dependent upon the load level, direction of forces applied and ligament integrity. We found that the ankle in dorsiflexion was more stable in the majority of the conditions tested. The plantarflexed ankle was the least stable. The articular geometry was the primary stabilizer of the ankle joint for simulated physiological loading. The lateral ankle ligaments were primary restraints in anterior translation, the deltoid ligament was most important in stabilizing posterior translation and both ligaments provided rotatory stability.

The present study represents a more critical analysis of the supporting elements of the ankle. We tested specimens under multiple loading conditions whereas some previous studies were conducted with a single loading condition. Previous studies did not specifically examine the effects of ankle position. Others examined rotatory stability but not translation stability. In the current study, we first determined the degree of laxity in the intact specimens, then used this limit in the testing of the injured conditions, in order to avoid damaging the remaining soft tissue constraints. In addition, the design of the testing device facilitated multi-axis testing without removing and repositioning the specimens repeatedly.

In previous studies, the range of applied forces to the ankle for translation tests varied from 50 N to 150 N (Hollis et al., 1995; Johnson and Markolf, 1983). Several of the previous biomechanical studies did not utilize the entire lower leg. In this study, the lower limb was

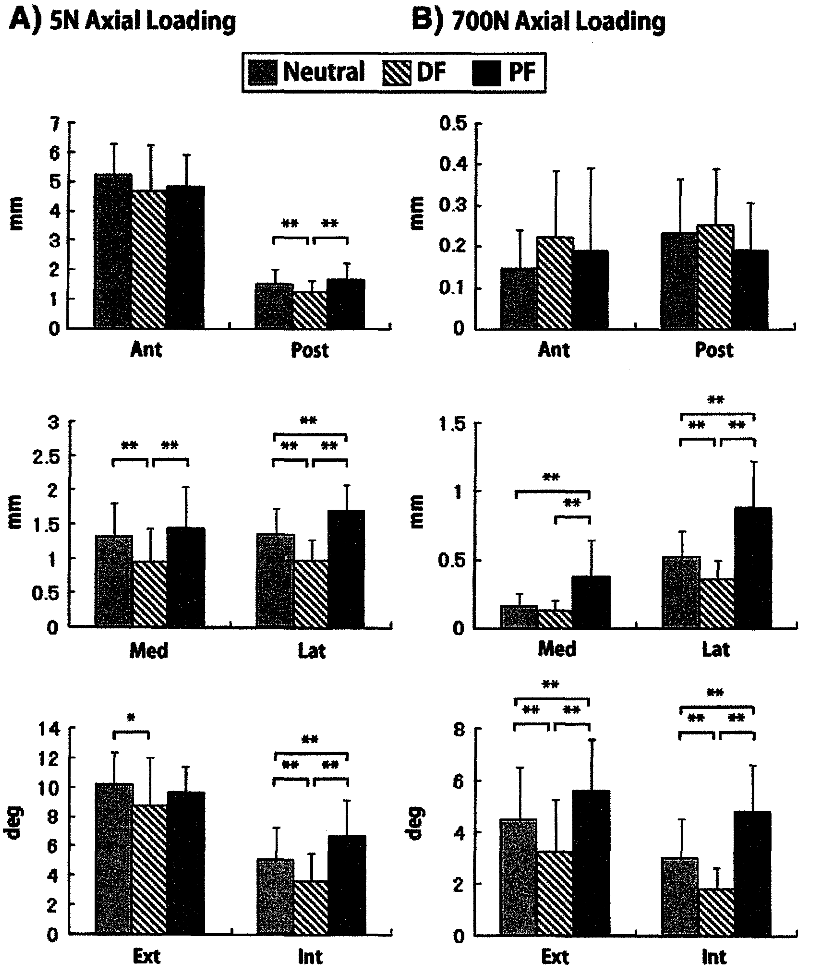

Fig. 3. Tibiotalar translation and rotation under A) 5 N axial load and B) 700 N axial load. Neutral, dorsiflexed (DF) and plantarflexed (PF) positions. (Statistically significant differences, *, $P<0.05$; **, $P<0.01$.)

utilized, maintaining motion between the tibia and fibula. Several studies reported that the fibula had apparent motion in all planes when loaded and therefore it is important to preserve the entire lower leg for testing (Barnett and Napier, 1952; Close, 1956; Stiehl, 1991). Recent investigations (Bahr et al., 1997; Cass and Settles, 1994; Hollis et al., 1995; Stiehl et al., 1993) applied these conditions to ankle stability tests and we believe they are more physiological.

Several investigators studied the range of anterior–posterior translation motion at the ankle. Hollis et al. (1995) reported 7.5 mm of anterior–posterior displacement for 50 N applied load in the neutral ankle position for anterior–posterior translation. Johnson and Markolf (1983) reported 6.6 mm of anterior–posterior displacement for 100 N and Bulucu et al. (1991) reported 6.4 mm of anterior displacement for 150 N. No axial load was applied to the ankle in any of these studies. These reported values of anterior–posterior translation were similar to our results. The effect of ankle position on anterior–posterior laxity measurements differed among investigators. Dorsiflexion was the most stable position in all three studies described above. However, Johnson and Markolf(1983) and Bulucu et al. (1991) found that the neutral position was associated with the greatest laxity, while Hollis et al. (1995) demonstrated greatest anterior–posterior laxity that in plantarflexion. In the current study, posterior displacement was significantly reduced in the dorsiflexed ankle compared to the neutral and plantar flexed positions.

There are few previous reports that studied medial–lateral translation at the ankle. In our study, the plantarflexed ankle position had significantly more medial–lateral motion than the dorsiflexed position. The geometry of the talus may be responsible for some of these differences. The width difference between the anterior (wider) and posterior (narrower) portions of the superior surface of the talar body was reported to average 2.4 mm (Stiehl, 1991). Our data reflected a significant difference in medial–lateral translation between plantar and dorsiflexed ankle positions. For internal–external rotation testing of the ankle, Johnson and Markolf (1983) reported a combined 13.8° of rotation at 2.5 Nm torque in neutral position with no axial load. McCullough and Burge (1980) reported a combined 24° for a 3 Nm torque limit with 9.8 N axial load. These values were consistent with our results of a combined 16.1° for 2.5 Nm at 5 N axial load for the neutral ankle position. Dorsiflexion was the most stable position for internal–external rotation in the current study, which was consistent with other reports (Johnson and Markolf, 1983; Stiehl et al., 1993).

Earlier reports investigated the biomechanics of lateral ligaments of the ankle. Both for *in vitro* and *in vivo* studies, rupture of the lateral ligament resulted in significant increases in anterior laxity (Bulucu et al., 1991; Glasgow et al., 1980; Hollis et al., 1995; Larsen, 1986). These findings supported our results in which the lateral ligaments contributed approximately 75% constraint to anterior translation. The deltoid ligament has not been investigated as thoroughly, especially for its effect on translational stability. Harper reported that the deltoid ligament was a secondary restraint against anterior talar translation when no axial load was applied (Harper, 1987, 1990). Our findings