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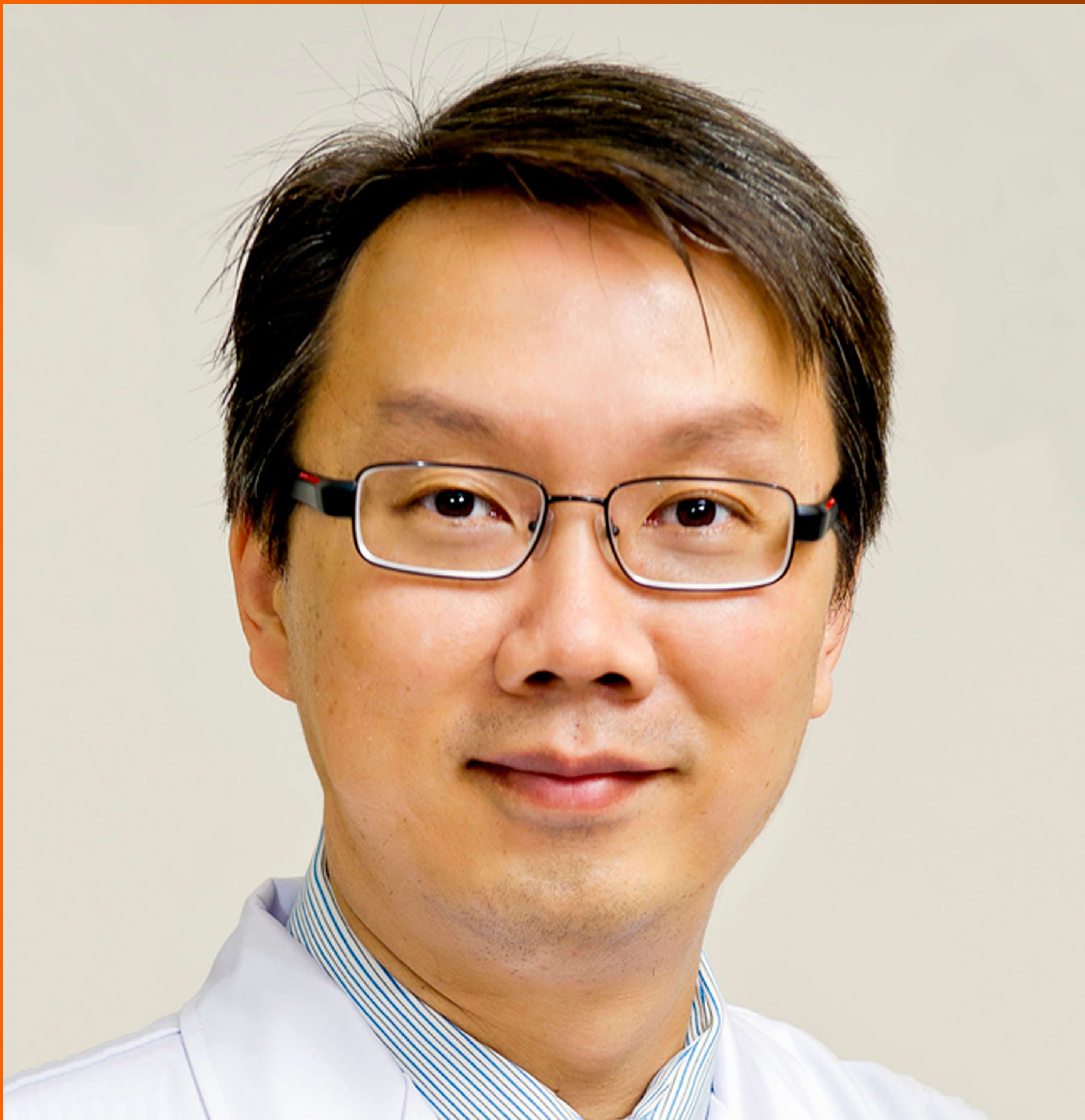

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MINIREVIEWS

Understanding the medial ulnar collateral ligament of the elbow: Review of native ligament anatomy and function

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Abstract

The medial ulnar collateral ligament complex of the elbow, which is comprised of the anterior bundle [AB, more formally referred to as the medial ulnar collateral ligament (MUCL)], posterior (PB), and transverse ligament, is commonly injured in overhead throwing athletes. Attenuation or rupture of the ligament results in valgus instability with variable clinical presentations. The AB or MUCL is the strongest component of the ligamentous complex and the primary restraint to valgus stress. It is also composed of two separate bands (anterior and posterior) that provide reciprocal function with the anterior band tight in extension, and the posterior band tight in flexion. In individuals who fail comprehensive non-operative treatment, surgical repair or reconstruction of the MUCL is commonly required to restore elbow function and stability. A comprehensive understanding of the anatomy and biomechanical properties of the MUCL is imperative to optimize reconstructive efforts, and to enhance clinical and radiographic outcomes. Our understanding of the native anatomy and biomechanics of the MUCL has evolved over time. The precise locations of the origin and insertion footprint centers guide surgeons in proper graft placement with relation to bony anatomic landmarks. In recent studies, the ulnar insertion of the MUCL is described as larger than previously thought, with the center of the footprint at varying distances relative to the ulnar ridge, joint line, or sublime tubercle. The purpose of this review is to consolidate and summarize the existing literature regarding the native anatomy, biomechanical, and clinical significance of the entire medial ulnar collateral ligament complex, including the MUCL (AB), PB, and

transverse ligament.

Key words: Elbow; Anterior bundle; Medial ulnar collateral ligament; Native anatomy; Biomechanics; Valgus stability

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Core tip: The anterior bundle of the medial ulnar collateral ligament complex plays a crucial role in elbow stability, specifically as a valgus and rotational constraint. Based on recent studies and our own cadaveric dissections, the ulnar footprint has a broader insertion that is more tapered and elongated than previous considered. A comprehensive understanding of the anatomy and biomechanical properties of the medial ulnar collateral ligament is imperative to optimize reconstructive efforts, and to enhance clinical and radiographic outcomes.

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INTRODUCTION

The medial ulnar collateral ligament [MUCL, also referred to as the ulnar collateral ligament (UCL), medial collateral ligament (MCL), and anterior bundle (AB)] is the primary restraint to valgus instability of the elbow[1-5]. The MUCL is one of three ligaments that comprise the "medial ulnar collateral ligament complex" of the elbow with the posterior bundle (PB) and transverse ligament (TL) being the other two (Figure 1). The MUCL, in particular, has been shown to be the primary stabilizer of the elbow during valgus stress, followed by the radial head and dynamic stabilizers of the elbow such as the flexor-pronator muscle mass[6-10]. The MUCL is composed of two separate bands (anterior and posterior) that provide reciprocal function with the anterior band tight in extension, and the posterior band tight in flexion.

The MUCL is commonly injured in overhead throwing athletes when a valgus moment is placed on the elbow during the late cocking and early acceleration phases[11-15]. Incompetence or rupture of the ligament leads to valgus instability which has varying clinical presentations. Patients may complain of instability, however, most will report pain, reduced accuracy, and decreased velocity. Clinically significant pathology often requires surgical intervention. Ligament reconstruction relies on appropriate graft positioning at both the humeral origin and the ulnar insertion. A thorough understanding the native anatomy of the MUCL facilitates the surgeon's ability to effectively restore stability and function.

In 1985, Morrey and An[10] published the first quantitative analysis of the medial ulnar collateral ligament complex. Based on 10 fresh frozen cadavers, they described the dimensions of the AB and PB at various degrees of elbow flexion. Since that time, multiple studies have assessed the anatomy and biomechanics of the AB[2,3,6,16,17]. Historically, general consensus held that the AB inserts solely and directly onto the sublime tubercle. However, recent studies have shown that the AB insertion is in fact broader, tapered, and significantly larger in terms of surface area than previously thought.

The purpose of this review is to consolidate and summarize the existing literature regarding the anatomy, biomechanical function, and clinical significance of the native (non-reconstruction) MUCL. It is our hope that this work may serve as a framework for better understanding valgus instability in the elbow and refining surgical techniques in order to optimize post-operative outcomes.

ANATOMY

As mentioned, the MUCL (AB) is the primary restraint to valgus instability of the elbow[1-5]. The PB is a soft tissue stabilizer of the elbow with contributions greatest during flexion[17]. It is generally thought that the TL does not provide a significant contribution to elbow stability[10]; however, recent study has revealed a direct insertion of the TL onto the AB that may potentially play a role in elbow stability[18]. The AB, in particular, has been shown to be the primary stabilizer of the elbow in valgus stress, with the radial head and dynamic stabilizers of the medial elbow also contributing[6-10]. It originates on the anterioinferior surface of the medial epicondyle and inserts onto the sublime tubercle of the ulna (Figure 1).

Origin

Surface area and footprint center: The origin of the AB is posterior to the elbow's axis of rotation, on the anterior, inferior, and lateral aspect of the medial epicondyle[10]. The surface area of the humeral origin has been widely variable in the literature (Table 1). Dugas *et al*[19] showed that the AB origin was round with a mean surface area of 45.5 mm^2 (range of 25.9-59.4 mm^2) in 13 fresh frozen cadavers utilizing a three-dimensional (3-D) electromagnetic tracking and digitalizing device. Similar to the findings of Dugas *et al*[19], a recent study by Camp *et al*[18] found a mean origin surface area of 32.3 mm^2. In contrast to these two studies, Frangiamore *et al*[20] found this measurement to be notably smaller at 17.0 mm^2 (range of 14.9-19.1 mm^2) through analysis of 10 fresh frozen cadaver specimens. Potentially due to differences in measurement techniques, these studies demonstrated variable results. Additionally, Frangiamore *et al*[20] and Dugas *et al*[19] described the center of the ligament origin in different terms, which is of clinical importance when determining the location of humeral tunnel placement during MUCL reconstruction surgery. Dugas *et al*[19] described the center of the origin as an area of tissue on a flat surface anterior and inferior to the medial epicondyle. The mean distance they measured was 13.4 mm from the center of the medial ep-

Table 1 Summary of anatomic studies describing the length, width, and surface area of the anterior bundle

Ref.	Specimen	AB length (mm)	AB width (mm)	Origin surface area (mm^2)	Insertion surface area (mm^2)	AB surface area (mm^2)
Alcid *et al*[2] 2004	-	27	4.5	-	-	121.5
Beckett *et al*[23] 2000	39	26.7	-	-	-	-
Camp *et al*[35] 2017	10	-		32.3	187.6	324.2
Dugas *et al*[19] 2007	13	-	6.8	45.5	127.8	-
Eygendaal *et al*[9] 2002	5	26	5	-	-	-
Farrow *et al*[22] 2014	10	53.9	-	-	-	-
Farrow *et al*[21] 2011	12	51.7	-	-	-	-
Floris *et al*[25] 1998	18	-	5.8	-	-	-
Morrey *et al*[10] 1985	10	27.1	4.7	-	-	-
Regan *et al*[17]	8	21.1	7.6	-	-	-
Safran *et al*[5] 2005	12	-	7.2	-	-	-
Timmerman *et al*[26] 1994	10	-	6	-	-	-
Frangiamore *et al*[20] 2018	10	21.5	-	17	66.4	-

AB: Anterior bundle.

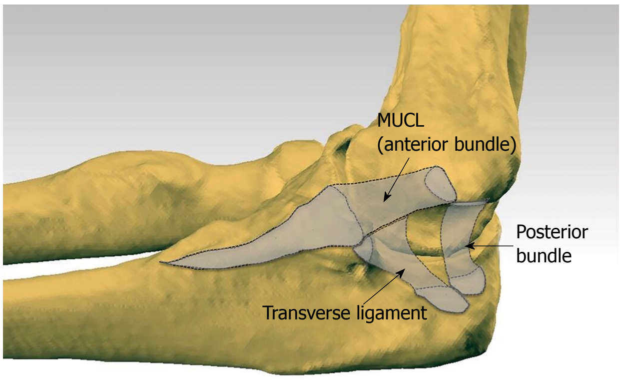

Figure 1 Medial ulnar collateral ligament complex of the elbow with outlined ligaments generated by co-registering the three dimensional digitized anatomy and computed tomography scan of a cadaveric elbow. Note the tapered and distally elongated insertion of the medial ulnar collateral ligament on the sublime tubercle and ulnar ridge. MUCL: Medial ulnar collateral ligament.

icondyle to the center of the origin, with Camp *et al*[18] finding a similar mean distance of 11.7 mm. Rather than describing this as a linear distance with no specific angle, Frangiamore *et al*[20] described this measurement in terms of two separate measurements, reporting the center of the origin to be located, on average, 8.5 mm distal (inferior) and 7.8 mm lateral (anterior) to the medial epicondyle. The use of two measurements relative to a single point in the latter study may assist with better reproducibility.

Insertion

Surface area and footprint center: Historically, the ulnar footprint of the AB has generally been described as inserting solely onto the sublime tubercle, serving as the anatomical landmark for surgical repairs and reconstructions. In one such early report, the mean AB insertional surface area was 66.4 mm^2[20].

Recently, authors have described the AB insertion as a longer, distally tapered area that follows the ulnar ridge. The surface area of this broader insertion has was reported by Dugas *et al*[19] to have a mean surface area of 127.8 mm^2. In this study, length of the ulnar footprint measured an average of 29.2 mm. Others have found similar lengths when appreciating a tapered insertion with means of 30.2 mm and 29.2 mm[21,22]. Further study by Camp *et al*[18] identified a tapered insertion with a mean surface area of 187.6 mm^2 and an ulnar footprint length averaging 29.7 mm.

Because the footprint center of a broader tapered insertion may not occur in the location previously assumed (apex of the sublime tubercle), the optimal position of the ulnar tunnel in reconstructive efforts may still need to be elucidated. Clinical relevance of the broader tapered ulnar insertion described in recent studies by Dugas *et al*[19] and Camp *et al*[18] is in need of further investigation.

One recent study of 10 cadaveric specimens has shown the mean proximal insertional width to be 9.4 mm which is greater than previously noted[18]. This discrepancy in widths leaves room for further investigation to ensure that the native anatomy of the AB is fully documented and understood (Figure 2).

Overall ligament dimensions

Length: The AB is the longest ligament of the medial elbow, spanning the inferior aspect of the medial epicondyle to the sublime tubercle and extending distally along the ulnar ridge. The average length of AB has been reported between 21.1 mm and 31.4 mm in several older studies[2,9,10,17,20,23]. These lengths were measured from the center of the origin to center of sublime tubercle, based on a direct insertion onto the sublime tubercle without a distal extension. In contrast, more recent reports measured from the center of the humeral origin to the most distal point of tapered insertion reported mean lengths of 53.9 mm[21] and 51.7 mm[22]. The difference in length measured between a non-tapered sublime insertion and a tapered insertion calls for further study

Table 2 Maximum physiologic valgus demonstrated in various studies with noted elbow position and load applied during testing

Authors	Specimens	Maximum valgus (degrees)	Elbow flexion (degrees)	Load (nm)
Callaway *et al*[3] 1997	28	3.6	30, 60, 90, 120	2
Floris *et al*[25] 1998	18	6	20	0.75
Morrey *et al*[4] 1991	6	5	20	Gravity
Safran *et al*[5] 2005	12	11.1	70	2

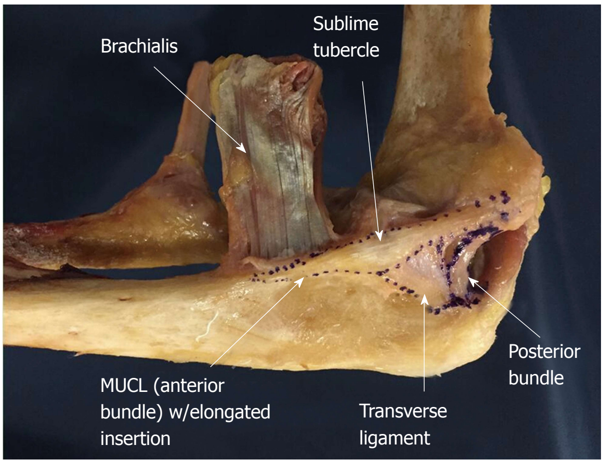

Figure 2 Cadaveric specimen outlining all ligaments of the medial ulnar collateral ligament complex including the medial ulnar collateral ligament/anterior bundle, posterior bundle, and transverse ligament. MUCL: Medial ulnar collateral ligament.

to evaluate native AB anatomy and the clinical relevance of different measurements. In particular, appropriate length of the ligament component has important implications for ligament reconstruction.

It is important to note that the AB is not an isometric soft tissue stabilizer but instead changes in length throughout flexion. Studies have shown the length of the AB changes by 18%, between 2.8 mm and 4.8 mm as the elbow moves from extension to flexion[6,10,24]. The dynamic length of the AB is an aspect of native biomechanics that must also be considered during reconstruction procedures.

Width: The width of the AB varies, increasing distally to its greatest width at the sublime tubercle before tapering to a point as it inserts distally along the ulnar ridge. Generally, there has been limited variability in the reported widths of the AB, ranging from 4.0 to 7.6 mm[2,5,10,17,25,26].

Surface area: The mean surface area of the AB has been reported between 108 mm^2 to 135 mm^2[2,10] in studies that did not take the full distal footprint into consideration. Given that Dugas *et al*[19] has shown the tapered ulnar footprint alone to have a mean surface area of 127.8 mm^2, the overall surface area of the ligament will undoubtedly be significantly greater than previously assumed in historical reports. In contrast to these historical reports, and in support of the Dugas *et al*[19] findings, a more recent study published by Camp *et al*[18] identified the mean surface area of the entire AB to be 324.2 mm^2 (Figure 3).

BIOMECHANICS

Valgus instability

The MUCL provides a vital contribution to the stability of the elbow when a valgus stress is applied. With an intact radial head, a fresh frozen cadaveric model demonstrated that the MUCL contributes 31% and 54% to valgus restraint at 0° and 90° of flexion, respectively[1]. In this study, 4 fresh frozen cadavers had a varying valgus load from 0 to 3 nm applied at 0° and 90° of flexion. With the maximum force applied, there was 3° of valgus laxity in full extension and 2° of laxity in flexion. Several studies have performed similar biomechanical testing at various degrees of flexion and various amount of force[2,3,5,6,10,25,27]. The amount of valgus laxity varies from 2° to 8° with an intact MUCL (Table 2).

At 30° and 90° of flexion, Callaway *et al*[3] showed valgus laxity of 3.6° under a 2nm load compared to an unloaded elbow. Safran *et al*[5] analyzed 12 cadaveric specimens with a 2Nm load applied at 30 degrees of elbow flexion, and showed a mean alignment of 10.7° of valgus with a neutral forearm rotation position. This study did not determine the inherent valgus alignment in an unloaded elbow, which affects the ability to compare these studies. Thus, with an intact MUCL and a 2 nm load applied, the amount of valgus laxity is generally greatest at 30° of flexion[5].

Other authors have evaluated the effect of transecting the AB on elbow stability (Table 3). When the AB is disrupted in cadaveric models, the amount of valgus instability increases until the point at which secondary osseous stabilizers such as the radial head impart stability. In the setting of AB deficiency, Callaway *et al*[3] demonstrated the greatest instability at 90° of flexion. The study reported a gain of 1.6, 2.8, 3.2, and 3.0 degrees of valgus motion at 30, 60, 90, and 120 degrees of flexion, respectively when compared to the intact state. Additionally, Mullen *et al*[28] showed that at 90 degrees of flexion, the transected AB increases valgus instability by 150%. Floris *et al*[25] and Søjbjerg *et al*[29] showed the greatest instability occurred at 70 degrees of flexion, with recorded valgus angles of 14.2 °and 11.8°. Safran *et al*[5] produced a maximal gain of 6.3°of laxity under 2 nm of valgus load with a transected AB at 50° of elbow flexion. Finally, Morrey *et al*[4] showed a gain of laxity over baseline ranging from

Table 3 Maximum valgus reported when the anterior bundle is transected in various studies with noted elbow position and load applied during testing

Ref.	Specimens	Maximum valgus gain (degrees)	Maximum absolute valgus	Elbow flexion (degrees)	Load (nm)
Ahmad *et al*[8] 2004	7	7.37	-	30	2
Callaway *et al*[3] 1997	28	3.2	-	90	2
Floris *et al*[25] 1998	18	5.7	-	30	0.75
Morrey *et al*[4] 1991	6	3.9	-	20	gravity
Safran *et al*[5] 2005	12	6.3	-	50	2
Sojbjerg *et al*[29] 1987	12	-	11.8	70	1.5

Some studies reported gain in valgus from physiologic state, while others noted absolute valgus.

Table 4 Summary of the studies evaluating the effect of the anterior bundle on relative internal rotation of the forearm in the native and pathologic state

Ref.	Specimen	Intact AB internal rotation (degrees)	Cut AB internal rotation (degrees)	Gain of internal rotation with cut AB
Bryce *et al*[6] 2008	-	4	-	-
Floris *et al*[25] 1998	18	6	18.5	12.5
Morrey *et al*[4] 1991	6	2.8	7.8	5

AB: Anterior bundle.

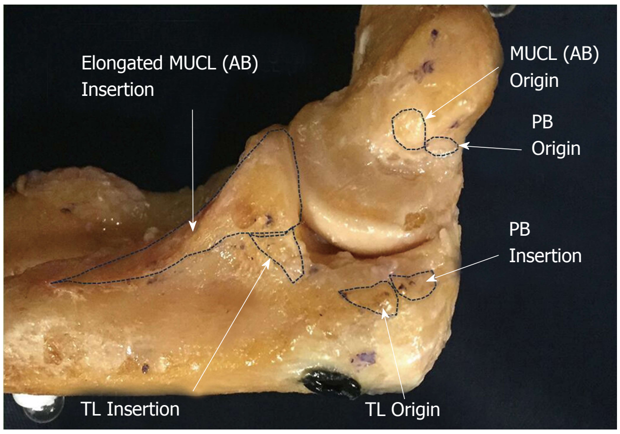

Figure 3 Cadaveric specimen showing origins and insertions of all ligaments of the medial ulnar collateral ligament complex including: Anterior bundle, posterior bundle and transverse ligament. AB: Anterior bundle; PB: Posterior bundle; TL: Transverse ligament.

3.3° to 4.8° in cadaver specimens at 20 degrees of flexion under gravity. In summary, an intact AB is vital in maintaining valgus stability of the elbow throughout the entire range of flexion.

Rotational instability

Internal rotation of the forearm relative to the humerus is constrained by the soft tissues stabilizer of the medial elbow. While there is inherent internal rotation during flexion, the degree of rotation is limited between 2.8°-6° in an uninjured elbow[4,6,25]. Transection of the AB permits internal rotation of the forearm to increase to 18.5° at 60° of joint flexion (Table 4)[25]. Correct understanding of these biomechanics is important in repair and reconstruction, as a rotatory moment is part of the mechanism of injury.

Tissue strength

Despite being the most frequently injured ligament in the overhead throwing athlete, the AB or MUCL has been shown to have the most inherent strength and stiffness[6,17]. This fact emphasizes the significant loads placed on the medial side of the elbow during the late cocking and early acceleration phase[30,31]. In a cadaveric model with each soft tissue stabilizer evaluated under stress, the AB was the strongest with an average load to failure of 260.9 N[17]. During overhead throwing, the elbow experiences 64 nm of mean valgus torque and 290 N of tensile force on the medial side (which is greater than the threshold for failure of 260.9 N)[30,31]. Furthermore, a maximal mean valgus load of 90Nm has been reported[32-34]. A recent study of 81 professional baseball pitchers (MLB and MiLB) over 82000 throws showed a mean valgus torque of 60 nm with individual participant means ranging from 41 nm to 94 nm[35]. Thus, it is clearly evident why the AB fails in this subset of athletes based on the load to failure being below the force imparted on the elbow.

CONCLUSION

The AB of the medial ulnar collateral ligament complex plays a crucial role in elbow stability, specifically as a valgus and rotational constraint. The AB originates on the humerus and inserts onto the sublime tubercle of the ulna. Based on recent studies and our own cadaveric dissections, the ulnar footprint has a broader insertion that is more tapered and elongated than previous considered. The data regarding the centers of the ulnar and humeral footprints provides guidance for proper tunnel placement during reconstructive efforts. The width and stiffness of the AB is described and can be used to guide graft selection during recons ruction.

Lastly, although the ligament is quite strong, the amount of force placed across the elbow in elite overhead throwing athletes is routinely exceeds the ligaments average load to failure. Accordingly, it is not surprising why MUCL injuries are so common amongst baseball players and pitchers.

Understanding native anatomy and biomechanics of the AB/MUCL is clinically important to help understand the pathoanatomy and guide surgical techniques when treating MUCL injuries.

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ORIGINAL ARTICLE

Basic Study

Comparison of a simplified skin pointer device compared with a skeletal marker for knee rotation laxity: A cadaveric study using a rotation-meter

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Author contributions: Puah KL performed the experiment, analysis and wrote the manuscript; Yew AKS designed and performed the experiment and edited the manuscript; Chou SM designed the experiment, obtained funding and edited the manuscript; Lie DTT designed the experiment, obtained funding and edited the manuscript.

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Abstract

AIM

To compare the measurements of knee rotation laxity by non-invasive skin pointer with a knee rotation jig in cadaveric knees against a skeletally mounted marker.

METHODS

Six pairs of cadaveric legs were mounted on a knee rotation jig. One Kirscher wire was driven into the tibial tubercle as a bone marker and a skin pointer was attached. Rotational forces of 3, 6 and 9 nm applied at 0°, 30°, 45°, 60° and 90° of knee flexion were analysed using the Pearson correlation coefficient and paired *t*-test.

RESULTS

Total rotation recorded with the skin pointer significantly correlated with the bone marker at 3 nm at 0° (skin pointer 23.9 ± 26.0° *vs* bone marker 16.3 ± 17.3°, r = 0.92; P = 0.0), 30° (41.7 ± 15.5° *vs* 33.1 ± 14.7°, r = 0.63; P = 0.037), 45° (49.0 ± 17.0° *vs* 40.3 ± 11.2°, r = 0.81; P = 0.002), 60° (45.7 ± 17.5° *vs* 34.7 ± 9.5°, r = 0.86; P = 0.001) and 90° (29.2 ± 10.9° *vs* 21.2 ± 6.8°, r = 0.69; P = 0.019) of knee flexion and 6 nm at 0° (51.1 ± 37.7° *vs* 38.6 ± 30.1°, r = 0.90; P = 0.0), 30° (64.6 ± 21.6° *vs* 54.3 ± 15.1°, r = 0.73; P = 0.011), 45° (67.7 ± 20.6° *vs* 55.5 ± 9.5°, r = 0.65; P = 0.029), 60° (62.9 ± 22.4° *vs* 45.8 ± 13.1°,

r = 0.65; P = 0.031) and 90° (43.6 ± 17.6° *vs* 31.0 ± 6.3°, r = 0.62; P = 0.043) of knee flexion and at 9 nm at 0° (69.7 ± 40.0° *vs* 55.6 ± 30.6°, r = 0.86; P = 0.001) and 60° (74.5 ± 27.6° *vs* 57.1 ± 11.5°, r = 0.77; P = 0.006). No statistically significant correlation with 9 nm at 30° (79.2 ± 25.1° *vs* 66.9 ± 15.4°, r = 0.59; P = 0.055), 45° (80.7 ± 24.7° *vs* 65.5 ± 11.2°, r = 0.51; P = 0.11) and 90° (54.7 ± 21.1° *vs* 39.4 ± 8.2°, r = 0.55; P = 0.079). We recognize that 9 nm of torque may be not tolerated *in vivo* due to pain. Knee rotation was at its maximum at 45° of knee flexion and increased with increasing torque.

CONCLUSION

The skin pointer and knee rotation jig can be a reliable and simple means of quantifying knee rotational laxity with future clinical application.

Key words: Rotatometer; Rottometer; Knee; Laxity; Cruciate; Biomechanics; Measurement

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Core tip: We describe a cadaveric study utilising a knee rotation jig paired with a skin pointer for the measurement of knee rotation laxity which has the potential for clinical application.

Puah KL, Yew AKS, Chou SM, Lie DTT. Comparison of a simplified skin pointer device compared with a skeletal marker for knee rotation laxity: A cadaveric study using a rotation-meter. *World J Orthop* 2018; 9(6): 85-91 Available from: URL: http://www.wjgnet.com/2218-5836/full/v9/i6/85.htm DOI: http://dx.doi.org/10.5312/wjo.v9.i6.85

INTRODUCTION

With increased interest in rotational stability with anterior cruciate ligament reconstruction as seen with the anatomical anterior cruciate ligament (ACL) reconstruction and the double-bundle ACL reconstruction, the need for an objective measurement of knee rotation arises in order to compare subjective clinical scores with rotational stability[1]. Registry data currently do not show any significant difference in knee outcome scores between single-bundle and double-bundle ACL reconstructions though proponents of the double-bundle technique recommend it as it is considered to be able to restore both rotational stability and anterior-posterior stability[1,2]. Stress radiography with the use of Roentgen Stereophotogrammetric Analysis (RSA) has been described previously with accuracy as high as 10-250 μm and 0.03-0.6° for translations and rotations, respectively, though it is an invasive procedure[3,4]. With variability of the pivot shift test amongst even trained orthopaedic surgeons, it becomes imperative that a non-invasive objective instrument be available to assess a patient's knee rotational stability[5]. There is a need for a portable, non-invasive yet simple to use device to measure knee rotation laxity in the clinic.

Almquist *et al*[6-8] has described a Rottometer which is a modified chair with the foot strapped to a rotating plate with measurements taken off a goniometer at the foot plate. However there was difference in the Rottometer readings compared to RSA at 90° flexion with 6 nm of torque and this has been attributed to be due to the measurements being taken at the foot which would thus include ankle rotation. To negate the effect of ankle rotation, we propose taking measurements off a fixed point more proximal and closer to the knee joint at the tibial tubercle with a non-invasive skin pointer while immobilizing the ankle in a foam boot. We designed a cadaveric study to assess the reliability of taking measurements off a non-invasive skin pointer placed over the tibial tubercle against that of a skeletally-mounted nail using a novel knee rotation jig modified from the Rottometer with a view to extending this to *in-vivo* testing.

To compare a non-invasive method of measuring knee rotation using a skin pointer against a nail fixed to the tibial tuberosity of a cadaveric knee specimen mounted on a knee rotation jig.

MATERIALS AND METHODS

Six pairs of cadaveric legs were mounted individually on a prototype knee rotation jig modified from the Rottometer described by Almquist *et al*[7] with a locking mechanism to set knee flexion at several predetermined flexion angles (Figure 1). These cadaveric legs were stored frozen and were thawed prior to use in this study. The jig, which is collapsible, foldable and portable, was securely clamped to a table using vice clamps. Each specimen was anchored to the jig at the femur with bolts for stability and at the foot and ankle with an Aircast® Foam Walker (Aircast, Summit, NJ, United States) attached to a rotating baseplate (Figure 2). The Aircast® Foam Walker was mounted to the baseplate to negate the effect of ankle rotation by immobilizing the ankle and foot. The jig features two adjustable metal side plates with Velcro straps which will be used for *in-vivo* testing subsequently. One Kirscher wire was driven into the apex of the tibial tubercle as a bone marker for reference and a skin pointer was attached above the tibial tubercle using a Velcro strap.

A torque wrench was attached to the baseplate and each knee was pre-conditioned prior to taking the first measurement against a mounted protractor. Using the torque wrench, a rotational force of 3, 6 and 9 nm was then applied at 0°, 30°, 45°, 60° and 90° of knee flexion. This was repeated 3 times at each torque and knee flexion for both internal and external rotation for each specimen. The respective readings of the bone marker and skin pointer were recorded and analysed using SPSS for Windows using the Pearson correlation coefficient and the paired *t*-test.

Table 1 Total knee rotation measured at 0°, 30°, 45°, 60° and 90° of knee flexion with 3 nm of torque

Knee flexion (°)	Total rotation (°)		Pearson's *r*	*r P*-value	*t*-test *P*-value
	Skin pointer	Nail			
0	23.88 ± 25.99	16.33 ± 17.32	0.92	0.000	0.032
30	41.70 ± 15.49	33.06 ± 14.66	0.63	0.037	0.042
45	48.97 ± 16.97	40.30 ± 11.20	0.81	0.002	0.030
60	45.73 ± 17.45	34.70 ± 9.45	0.86	0.001	0.008
90	29.21 ± 10.89	21.15 ± 6.75	0.69	0.019	0.016

Figure 1 Knee rotation jig prior to mounting of specimen.

Figure 2 Knee rotation jig with specimen mounted with nail though tibial tuberosity and skin pointer in place.

RESULTS

The readings for total rotation obtained with the skin pointer significantly correlated with that of the bone marker at 3 nm at 0°, 30°, 45°, 60°, and 90° of knee flexion (Table 1 and Figure 3). Similarly the readings for total rotation obtained with the skin pointer significantly correlated with that of the bone marker at 6 nm at 0°, 30°, 45°, 60° and 90° of knee flexion (Table 2 and Figure 4). However, although the readings between the skin pointer and bone marker correlated significantly at 3 nm of torque, there was a significant difference on paired *t*-test between the two readings through all degrees of flexion. With 6 nm of torque, there was a significant difference between the readings at 45°, 60° and 90° of flexion.

With 9 nm of torque, there was a statistically significant correlation at 0° and 60° but no statistically significant correlation at 30° and 90° of knee flexion though there was a similar trend to 3 and 6 nm of torque (Table 3 and Figure 5). With 9 nm of torque, there was a significant difference between the readings at 45°, 60° and 90° of flexion. We found that at 9 nm torque, the cadaveric specimen would not return to the neutral starting position, suggestive of deformation of the specimen.

The skin pointer exaggerated the amount of rotation compared to the bone marker at all torques and angles of knee flexion with the maximum difference of 15.6° at 45° knee flexion with 9 nm of torque. For both the skin pointer and the bone marker, knee rotation increased with increasing knee flexion with maximum rotation at 45° flexion with subsequent decrease in rotation till 90° of knee flexion was reached (Figures 3-5). With increasing torque at a fixed flexion, knee rotation increased (Figure 6).

DISCUSSION

Apart from stress radiography with the use of RSA which is an invasive procedure, other instruments have been described to measure knee rotation including Almquist's Rottometer from which our prototype jig is based on, Lars Rotational Laxiometer, Vermont Knee Laxity Device, Tsai *et al*[9]'s rotational knee laxity measurement device and Ahrens' torsiometer[7,10-12].

Almquist's Rottometer includes a chair where measurements were taken from the foot which may have contributed to its reported inaccuracy as ankle and foot rotation could still contribute to movement and readings[6,7]. The use of an Aircast® Foam Walker boot to immobilize the foot and ankle and the use of a skin pointer close to the knee joint as in our study would help to minimize systematic error from foot and ankle movement. Mouton *et al*[13,14] used a prototype rottometer with a similar ski boot and delivered the torque through a rotational handle bar and measured rotation through an inclinometer attached to the bar. Tsai's device utilized a magnetic tracking system with an Aircast® Foam Walker boot with reliable results[9]. Ahrens' utilized a torsiometer with Schanz pins to mount the cadaveric limbs skeletally with a potentiometer to measure rotation and demonstrated that cadaveric knees with arthroscopically resected ACLs had greater rotation than cadaveric knees

Table 2 Total knee rotation measured at 0°, 30°, 45°, 60° and 90° of knee flexion with 6 nm of torque

Knee flexion (°)	Total rotation (°)		Pearson's *r*	*r P*-value	*t*-test *P*-value
	Skin pointer	Nail			
0	51.12 ± 37.73	38.61 ± 30.07	0.90	0	0.064
30	64.64 ± 21.61	54.27 ± 15.11	0.73	0.011	0.051
45	67.73 ± 20.60	55.48 ± 9.45	0.65	0.029	0.019
60	62.85 ± 22.43	45.79 ± 13.05	0.65	0.031	0.006
90	43.61 ± 17.56	30.97 ± 6.25	0.62	0.043	0.007

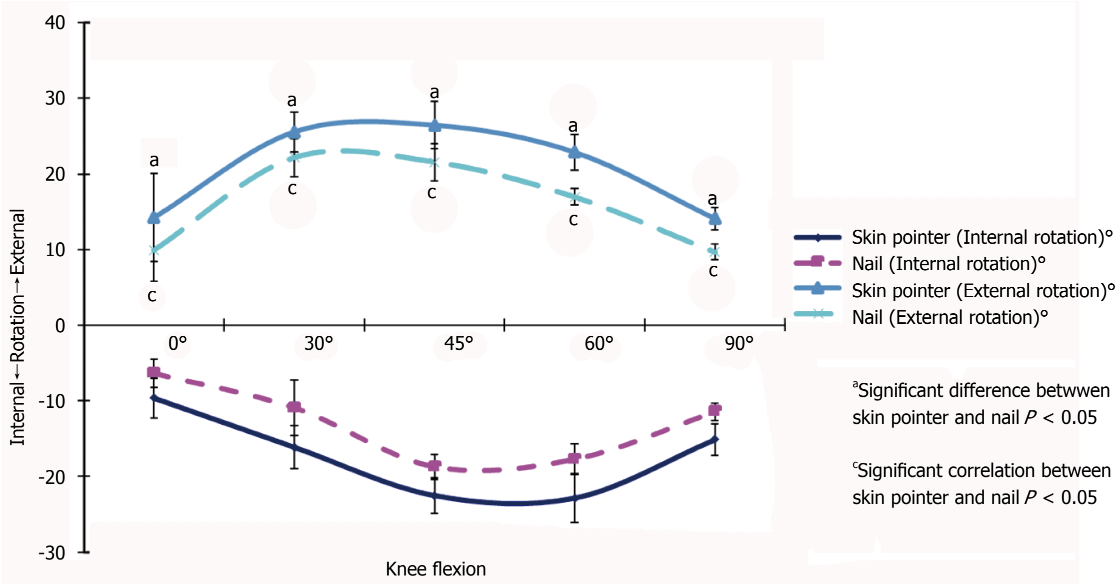

Figure 3 Knee rotation measured at 0°, 30°, 45°, 60° and 90° of knee flexion with 3 nm of torque.

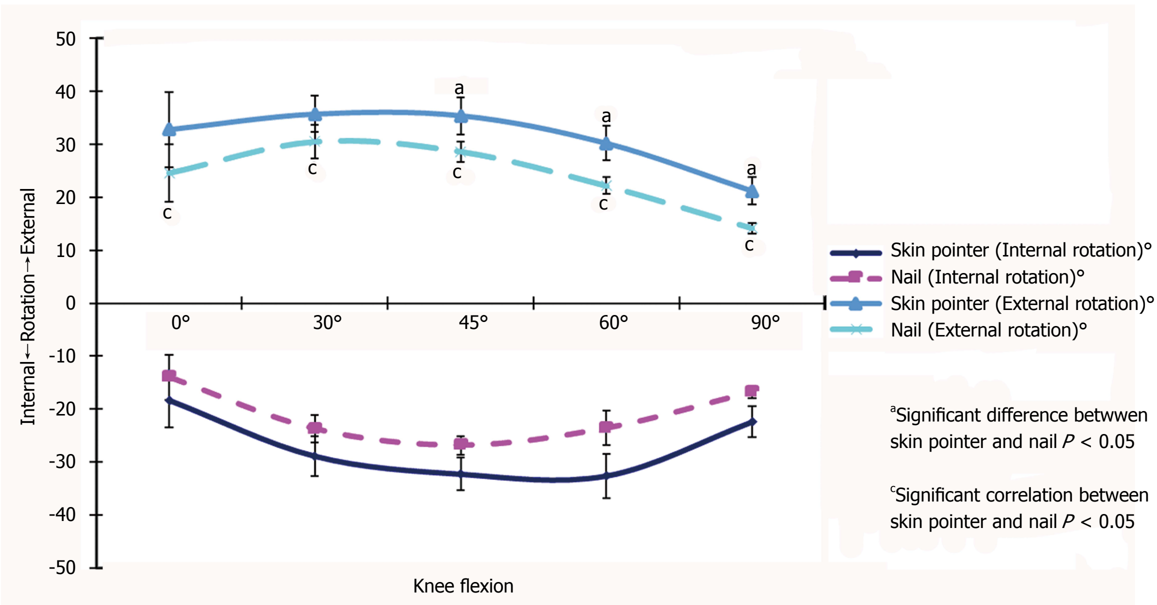

Figure 4 Knee rotation measured at 0°, 30°, 45°, 60° and 90° of knee flexion with 6 nm of torque.

with the ACL intact[12].

Robotic arm technology has also been described to deliver the rotational force to mimic the dial test[15,16]. The Rotab® device measures medial knee rotation when delivering an anterior translation force to measure anteromedial knee instablility[17]. A similar device which measures passive medial knee rotation with anterior translation of the tibia was described by Kurimura *et al*[18].

Table 3 Total knee rotation measured at 0°, 30°, 45°, 60° and 90° of knee flexion with 9 nm of torque

Knee flexion (°)	Total rotation (°)		Pearson's *r*	*r P*-value	*t*-test *P*-value
	Skin pointer	Nail			
0	69.67 ± 39.91	55.61 ± 30.61	0.86	0.001	0.046
30	79.18 ± 25.14	66.91 ± 15.42	0.59	0.055	0.072
45	80.67 ± 24.65	65.48 ± 11.23	0.51	0.112	0.040
60	74.52 ± 27.57	57.09 ± 11.50	0.77	0.006	0.017
90	54.70 ± 21.05	39.39 ± 8.22	0.55	0.079	0.018

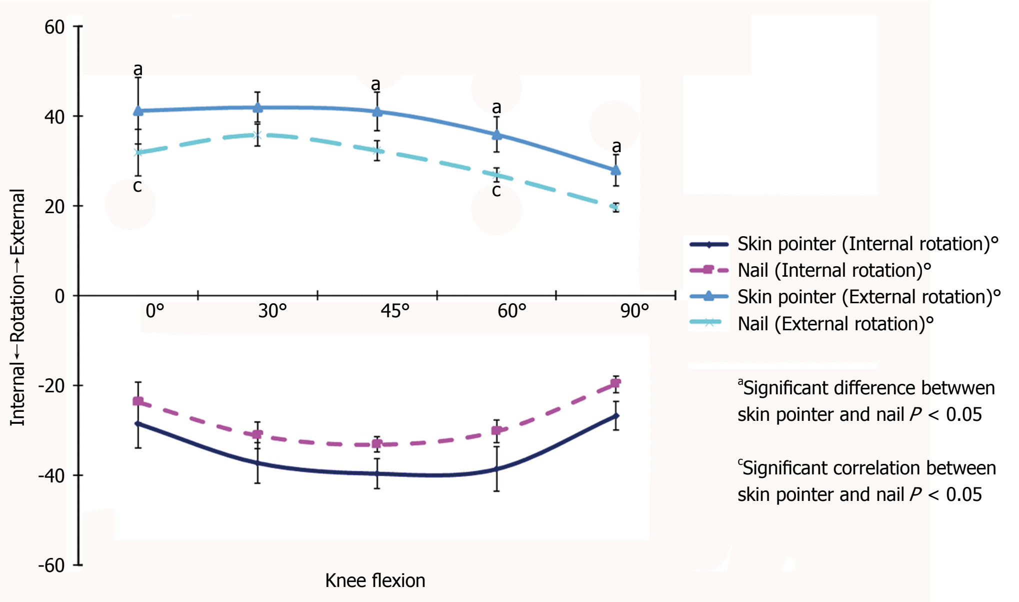

Figure 5 Knee rotation measured at 0°, 30°, 45°, 60° and 90° of knee flexion with 9 nm of torque.

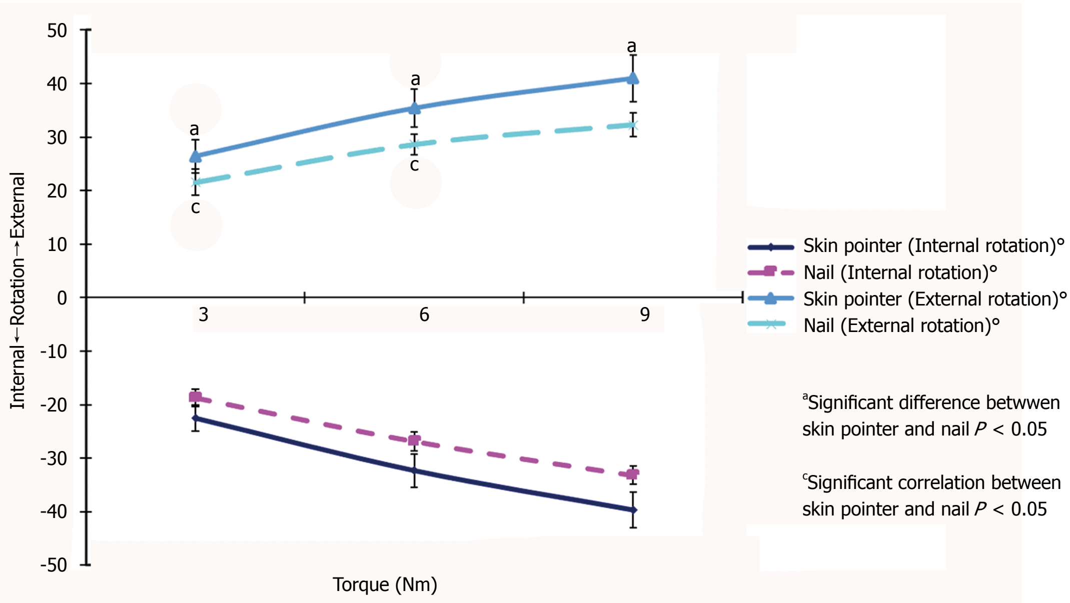

Figure 6 Knee rotation at 45° of flexion with 3, 6, and 9 nm of torque.

Hoshino *et al*[19] described a motion capture method using skin markers to measure the anterior translation of the distal femur in anaesthetised patients undergoing the pivot shift test.

Computer assisted surgery (CAS) devices which use motion-tracking technology and bony reference points can be used too but are invasive and are best used in the operating theatre during surgery[20,21]. The benefit of our setup is that it is simple to set-up, portable and does not rely on bulky electronic equipment allowing it to be used in the clinic and possibly at sports training grounds where a controlled environment with a ready electrical source may not be available.

Our study shows that our simple non-invasive skin pointer used in combination with our knee rotation jig can measure knee rotation similar to that of a skeletally placed marker with our knee rotation jig. We recognize that the skin pointer would exaggerate the amount of rotation compared to the bone marker due to movement of the soft tissue and skin overlying the bone. Hence although the readings between the skin pointer and the bone marker were significantly correlated with a similar trend, with higher torque and greater knee flexion, there were significant different between the individual measurements. We recognize too that 9 nm of torque may be not be well-tolerated in live human subjects due to pain. The effect of soft tissue causing an exaggeration of results has been reported previously[22]. Furthermore, 9 nm of torque may have caused deformation of our specimens affecting our results as a limitation of our use of cadavers which we may not observe *in vivo*.

Our jig and measurements assume a global single-axis of rotation of the knee, not taking into account translation of the knee which may occur *in vivo* with live subjects where rigid skeletal mounting to the jig will not be feasible. Hence off-axis movements may not be accurately measured as compared to a system where measurements are taken at both the femur and tibia taking into account movement of the subject in the jig. We recognize that the rotation axis of the tibia changes with knee as reported by Matsumoto and that our simple all-mechanical jig and measurement system may not be able to account for this change in axis[23]. Similar to Matsumoto's study, we found that the magnitude of knee rotation increases as the knee is flexed which then decreases as flexion reaches 90°. Knee rotation was observed to be at its maximum at 45° of knee flexion and it increased with increasing torque. Similar to other previously described instruments, our device measures rotation without the effect of weight-bearing[7,9-11].

Our study compared the use of a skin pointer in combination with our knee rotation jig against a skeletally mounted marker which showed significant correlation between the two readings. We recognise the significant difference between the absolute values of the two different measurement methods due to soft tissue movement over the bone and that the soft tissue in a cadaver which has been frozen and thawed will have different properties compared to that of a live subject. Similar experiments with Rotatometers/Rottometers using only live subjects with no skeletally mounted reference for comparison demonstrate high inter- and intra-observer reliability[24,25].

Objective measurement for anterior-posterior laxity using the KT-1000 Arthrometer is well accepted[26-28]. Our aim is to eventually develop a portable and user-friendly device analogous to the KT-1000 which can be used for objective measurement of knee rotation in a non-invasive manner. The investigation of the utility of our rotation jig mated with a robotic arm for kinematic measurements is currently ongoing which may negate the effect of translation of the knee *in vivo*. Our next phase is to collect data on volunteers with uninjured knees followed by patients with knee injuries and patients after surgery to document changes in knee rotational laxity with pathology and treatment.

In conclusion, the skin pointer combined with a knee rotation jig can be a reliable and simple means of quantifying knee rotation in the cadaveric knee with potential application *in vivo* as a non-invasive means of measuring knee rotation in the clinic.

ARTICLE HIGHLIGHTS

Research background

With double-bundle and anatomical single-bundle anterior cruciate ligament reconstruction for restoration of rotational knee kinematics, the need for objective clinical measurement of knee rotational laxity arises. Evaluation of knee rotation remains a challenge with intra-observer variability in the pivot shift test.

Research motivation

We aim to compare a non-invasive skin pointer with a knee rotation jig in cadaveric knees against a skeletally mounted marker.

Research methods

Six pairs of cadaveric legs were mounted on a knee rotation jig. One Kirscher wire was driven into the tibial tubercle as a bone marker and a skin pointer was attached. Rotational forces of 3, 6 and 9 nm applied at 0°, 30°, 45°, 60° and 90° of knee flexion. Results were analysed using the Pearson correlation coefficient and paired *t*-test.

Research results

Total rotation recorded with the skin pointer significantly correlated with the bone marker at 3 nm at 0° (skin pointer 23.9 ± 26.0° *vs* bone marker 16.3 ± 17.3°, r = 0.92; P = 0.0), 30° (41.7 ± 15.5° *vs* 33.1 ± 14.7°, r = 0.63; P = 0.037), 45° (49.0 ± 17.0° *vs* 40.3 ± 11.2°, r = 0.81; P = 0.002), 60° (45.7 ± 17.5° *vs* 34.7 ± 9.5°, r = 0.86; P = 0.001) and 90° (29.2 ± 10.9° *vs* 21.2 ± 6.8°, r = 0.69; P = 0.019) of knee flexion and 6 nm at 0° (51.1 ± 37.7° *vs* 38.6 ± 30.1°, r = 0.90; P = 0.0), 30° (64.6 ± 21.6° *vs* 54.3 ± 15.1°, r = 0.73; P = 0.011), 45° (67.7 ± 20.6° *vs* 55.5 ± 9.5°, r = 0.65; P = 0.029), 60° (62.9 ± 22.4° *vs* 45.8 ± 13.1°, r = 0.65; P = 0.031) and 90° (43.6 ± 17.6° *vs* 31.0 ± 6.3°, r = 0.62; P = 0.043) of knee flexion and at 9 nm at 0° (69.7 ± 40.0° *vs* 55.6 ± 30.6°, r = 0.86; P = 0.001) and 60° (74.5 ± 27.6° *vs* 57.1 ± 11.5°, r = 0.77; P = 0.006). No statistically significant correlation with 9 nm at 30° (79.2 ± 25.1° *vs* 66.9 ± 15.4°, r = 0.59; P = 0.055), 45° (80.7 ± 24.7° *vs* 65.5 ± 11.2°, r = 0.51; P = 0.11) and 90° (54.7 ± 21.1° *vs* 39.4 ± 8.2°, r = 0.55; P = 0.079). We recognize that 9 nm of torque may be not tolerated *in vivo* due to pain.

Research conclusions

We have measured knee rotation on a cadaveric knee utilising a knee rotation jig paired with a skin pointer against that of a skeletally mounted bone marker and have found a significant correlation between the two methods for the same magnitude of torque and knee flexion. We recognise that the use of the skin pointer introduces error due to movement of the soft tissue which increases with increasing torque.

Research perspectives

Our aim is to eventually develop a portable and user-friendly device which can be used for objective measurement of knee rotation laxity in a non-invasive manner. This may entail the use of accelerometers or robotic arms to measure kinematics.

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