Journal of Biomechanics 48 (2015) 2227–2231

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Short communication

Mid-range shoulder instability modeled as a cam-follower mechanism

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ARTICLE INFO

Article history:
Accepted 28 February 2015

Keywords:
Glenohumeral instability
Kinematics
Modeling

ABSTRACT

In this paper, we model a simplified glenohumeral joint as a cam-follower mechanism during experimental simulated dislocation. Thus, humeral head trajectory and translational forces are predicted using only contact surface geometry and compressive forces as function inputs. We demonstrate this new interpretation of glenohumeral stability and verify the accuracy of the method by physically testing a custom-molded, idealized shoulder model and comparing data to the output of the 2D mathematical model. Comparison of translational forces between experimental and mathematical approaches resulted in r^2 of 0.88 and 0.90 for the small and large humeral head sizes, respectively. Comparison of the lateral displacement resulted in r^2 of 0.99 and 0.98 for the small and larger humeral head sizes, respectively. Comparing translational forces between experiments and the mathematical model when varying the compressive force to 30 N, 60 N, and 90 N resulted in r^2 of 0.90, 0.82, and 0.89, respectively. The preliminary success of this study is motivation to introduce the effects of soft tissue such as cartilage and validation with a cadaver model. The use of simple mathematical models such as this aid in the set-up and understanding of experiments in stability research and avoid unnecessary depletion of cadaveric resources.

1. Introduction

The shoulder joint exhibits ball-and-socket-like kinematics with limited humeral head (HH) translation (Graichen et al., 2000; Howell and Galinat, 1989; Kelkar et al., 2001; Poppen and Walker, 1975). Shoulder instability is defined as the inability to maintain HH centering within the glenoid fossa (Graichen et al., 2000; Howell and Galinat, 1989; Lippitt et al., 1993; Matsen et al., 1991; Poppen and Walker, 1975). Biomechanically speaking, instability is a condition defined by the inability of a joint to return to its original position after perturbation (Kelkar et al., 2001; Leipholz, 1987). Clinically, the perturbed configuration corresponds to dislocation, while instability refers to larger than acceptable translation occurring during force exertion (Veeger and van der Helm, 2007).

In midrange shoulder joint motion, concavity-compression is the most important shoulder joint stability mechanism (Lazarus et al., 1996; Lippitt et al., 1993). Concavity-compression is the mechanism in which the convex humeral head is stabilized against translating forces by the concave glenoid fossa. Stability is related to the compression magnitude (Fukuda et al., 1988) and the constraint offered by the glenoid concavity (Anglin et al., 2000). Experimental studies investigating shoulder stability have either measured translational force during driven displacement (Favre et al., 2011; Halder et al., 2001; Itoi et al., 2000; Karduna et al., 1997; Kikuchi et al., 2008; Tammachote et al., 2007; Yamamoto et al., 2009) or have measured displacement after applying a predetermined translational force (Bryce et al., 2010; Elkinson et al., 2012; Giles et al., 2011; Mihata et al., 2012; Peltier et al., 2012; Schulze-Borges et al., 2013; Sekiya et al., 2012, 2009; Wellmann et al., 2011a, 2011b, 2008). Oosterom et al. (2003) published a mathematical model predicting the "translational stiffness" of the shoulder joint after arthroplasty. The authors define a mathematical model for glenohumeral prostheses defining translation force over 3 distinct "positions" and 2 "regions" of instability (Fig. 1). Briefly, from the centered position (position 1) the HH slides (region 2) until the sliding angle, ϕ_1, is equal to the constraint angle, θ, at the glenoid rim (position 3). Sliding beyond position 3 (region 4) is termed subluxation, until the subluxation angle, ϕ_2, is equal to θ, at which point the HH is truly dislocated (position 5). Total travel along the translation direction until position 5 is reached is commonly referred to as "distance to dislocation".

Recently, much attention has focused on the role of bone loss in glenohumeral instability. Clinical and experimental data suggests that both glenoid and humeral bone loss may significantly increase recurrent instability risk (Kaar et al., 2010; Walia et al., 2012).

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http://dx.doi.org/10.1016/j.jbiomech.2015.02.053
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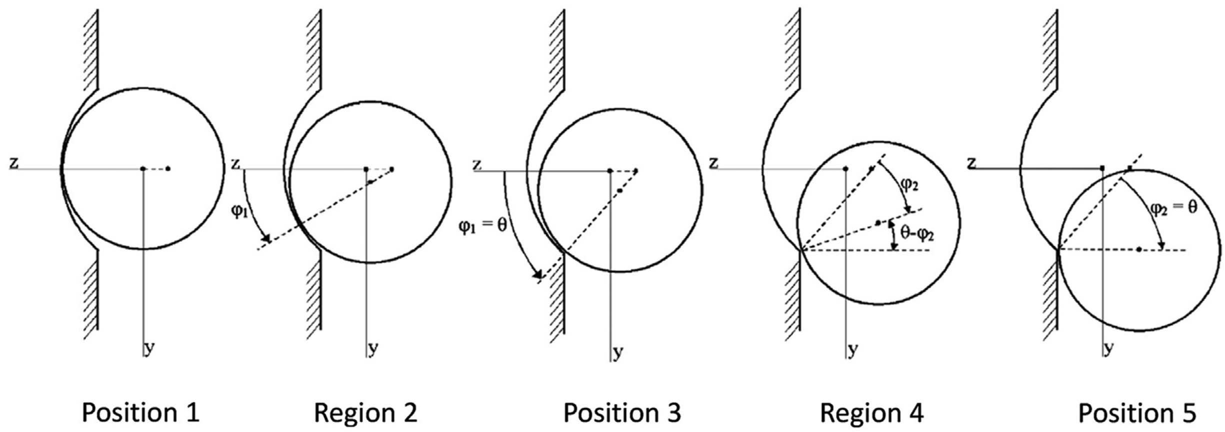

Fig. 1. Regions and change positions of glenhumeral motion decribed by Oosterom et al. θ=glenoid component constraint angle, calculated by glenoid chord/2 radius of curvature of the glenoid, $\Phi 1$=instantaneous angle between the z-axis, origin of the glenoid sphere and the instantaneous center of HH, $\Phi 2$=instantaneous angle between the origin of the glenoid sphere, the glenoid rim and the instantaneous center of HH. Used with permission of Elsevier. Previously published in Oosterom et al. (2003).

Theoretically, the glenoid track concept illustrates the mechanism by which bone loss on either part of the joint interferes with glenohumeral stability due to altered articular geometry (Trivedi et al., 2014; Yamamoto et al., 2007).

Our hypothesis is that the kinetic and kinematic interactions within the glenohumeral joint during simulated experimental dislocation (translation under compression), can be accurately defined with machine design principles, modeling the joint as a cam-follower mechanism. From this perspective, the HH functions as the follower translating over the cam-like glenoid surface. Thus, humeral head trajectory and translational forces could be predicted using only contact surface geometry and compressive forces as function inputs. Compared to previous work by Oosterom et al. (2003), we derived equations valid for non-circular shapes, making the concept applicable to a variety of clinical purposes such as the study of native glenohumeral anatomy and the influence of bone defects on joint stability. We illustrate this new method of determining glenohumeral instability and characterize the accuracy of the approach by physically testing an idealized shoulder model and comparing data to the output of 2D mathematical models. We also evaluate model robustness by examining interactions with both small and large HH's under different compressive loads.

2. Materials and methods

2.1. Experimental set-up

An idealized physical model of a ball and socket, approximating a humeral head and a glenoid, was fabricated with careful control over geometric parameters. The glenoid was modeled using semi-rigid urethane casting resin (Smooth-Cast 65®D, Smooth-On, Inc, Easton, PA, USA) cast over a 80-mm diameter sphere to form a concave articulating surface, resulting in an axisymmetric cup 39.6 mm in radius and 17.5 mm deep with a constraint angle, θ, of 56°. Two HH sizes were created using off-the-shelf polyoxymethelene spheres, one small (25.4-mm radius) and the other large (38.1-mm radius). Heads were fitted onto a 175-mm long polycarbonate rod simulating the humeral shaft. The marked difference in humeral head sizes was deliberately chosen to accentuate differences in experimental test results. Although deviating head sizes limits the anatomical accuracy of the glenohumeral model, they do not compromise the aim of this paper to characterize the performance of the mathematical model.

The models were mounted onto a custom-built four-axis electromechanical set-up, its use previously reported in cadaver experiments assessing midrange glenohumeral stability (Halder et al., 2001; Itoi et al., 2000; Matsuhashi et al., 2013; Yamamoto et al., 2009, 2010, 2012). In the first experiment the stability of small and large HH models was evaluated under 50-N compressive force applied during translation. In the second experiment the small HH was translated under 30 N, 60 N and 90 N compressive loads, consecutively. Silicone lubricant was sprayed on articulating surfaces before each test. Translation was initiated from a position 50 mm ahead of the most medial (reference) point and progressed under displacement control at 2 mm/s for 100 mm along the y-axis. Position and force data collection was initiated after the HH crossed the reference point.

2.2. Mathematical 2D cam-follower model

In this mathematical approach to analytically assess joint stability, the glenohumeral joint was modeled with a spherical follower emulating the HH and a linear translating cam emulating the glenoid, as shown in Fig. 2. The follower was assumed to be circular with radius, r. The cam surface is represented as a function,$f(y)$. The follower trajectory,$g(y)$, can be found by simple addition of position vectors, one between follower center and the instantaneous contact point, $\bar{r}^*_{C/H}$, and the second defining the contact point relative to the most medial position reference, $\bar{r}^*$, (Eq. (1)). Resolving this vector sum into components, expressions for y and $g(y)$ relative to the contact point are obtained (Eq. (2)). The instantaneous slope of the cam profile can be calculated differentiating the cam profile, $(d/dy)f(y)$ (Eq. (3)). Substituting the expression for tangent angle, β, and applying a shift in y using the relationship in Eq. (2), a general solution for the follower, or HH, path is obtained (Eq. (4)). Similar expressions could be developed for an irregularly shaped humeral head with surface contour $h(y)$; however, numerical methods would likely be required to obtain solutions. If frictionless contact between the HH and glenoid surface is assumed, the force required to translate the HH,$F_Y(y)$, for a constant axial load, F_Z, can be determined from static equilibrium equations (Eq. (5)). Results are reported as the ratio between $F_Y(y)$ and F_Z to allow for extrapolation to values commonly reported in the literature.

$$\bar{r} = \bar{r}^* - \bar{r}^*_{C/H} \tag{1}$$

$$y = y^* - r\cdot\ \sin\left(\beta(y^*)\right) \qquad g(y) = f\ (y^*) + r\ \cos\left(\theta(y^*)\right) \tag{2}$$

$$\theta(y) = \arctan\left(\frac{d}{dy}f(y)\right) \tag{3}$$

$$g\ (y - r\cdot\ \sin\ (\beta\ (y)\ )\ ) = f\ (y) + r\cdot\ \cos\ (\beta\ (y)\ ) \tag{4}$$

$$F_Y(y - r\cdot\ \sin(\beta\ (y)\ )\ ) = F_Z\cdot\ \tan\ (\beta\ (y)\ ) \tag{5}$$

If the cam path is circular within region 2, as described previously by Oosterom and detailed in Fig. 3(left), equations for the HH path (Eq. (6)) and translation force (Eq. (7)) are greatly simplified, with the center moving along a circular path of radius $(\rho - r)$.

$$g(y) = -\sqrt{(\rho - r)^2 - y^2} + (\rho - r) \tag{6}$$

$$F_y(y) = Fz\frac{y}{\sqrt{(\rho - r)^2 - y^2}} \tag{7}$$

After reaching articular position 3 (Fig. 1), the follower center is described by another circular path with radius of curvature, r(Eq. (8)). As detailed in Fig. 3 (right), translation force for region 4 can be expressed by Eq. (9): (equation adapted from Oosterom et al.)

$$g(y) = -\sqrt{r^2 - (y - l)^2} + (d - r) \tag{8}$$