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IMPACT FORCE IN SHOULDER DISLOCATION TRAUMA IN TEAM SPORTS

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ABSTRACT. This scientific paper presents the biomechanical aspects of the kinetics of the collision of two athletes, with consequences on the traumatism of shoulder dislocation, as well as the ways to reduce the impact force to avoid injury. Using the value of the coefficient of restitution upon collision, based on previous own research, the analytical expression of the impact force is determined, this depending on the instantaneous velocities before and after the collision, on the coefficient of restitution and on the masses of the two athletes. Shoulder dislocation occurs due to the force acting on the humeral head and the joint dislocation from the glenoid cavity of the scapula, this force being the impact force upon collision. Using the game of rugby as an example, the impact forces are calculated for several collisions during a sports competition and then compared with the joint resistance force above which the shoulder joint dislocation can occur. This last force is determined taking into account the mechanical rupture stresses of the articular ligaments, muscles, joint capsule and bone tissue. The paper presents a numerical simulation for determining the pre-collision velocities, for different masses of athletes, so that posterior shoulder dislocation occurs. Also, the paper numerically simulates various situations for the joint resistance force that keeps the scapula-humeral joint in normal operating conditions and then compares this with the impact force from the rugby game.

Keywords: joint dislocation, collision force, rugby, prevention, orthosis.

INTRODUCTION

The accidents can be prevented if the player knows details about dynamic impact, impact techniques, how to protect himself during impacts. Thus, there was analyzed the dependence between the frequency and intensity of the impact

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and cervical accidents, for the orderly pile in the game of rugby (Scher, 1987; Milburn, 1993; Quarrie, 2001; Yeomans, 2018), the spinal column accidents (Silver, 1988, 1993, 1994) or knee accidents (Ellapenl, 2016) due to impact during the game.

Shoulder dislocation resulting from accidents in team sports has an immediate effect of incapacitating the athlete for a period, depending on the severity of the trauma (without or with ligament or muscle rupture). The duration of immobilization depends on the type of dislocation, the patient's age, whether it is the first dislocation or a relapse, the level of sports activity, etc. (Cunningham, 2005; Radhik, 2022). An epidemiological study conducted among young athletes in the USA (Twomey-Kozak, 2021) highlighted the fact that most shoulder injuries occur among young athletes, usually amateurs, who have little experience with injuries. Over a five-year period, around 89,500 shoulder dislocations were reported, with injuries occurring in sports such as basketball (24.1%), American football (21%), soccer (7.1%), baseball (7%), etc.

The purpose of this paper is to present a theoretical biomechanical model for evaluating the maximum resultant force given by the muscles and ligaments in the glenohumeral joint and to compare this force, for a human subject, with the values obtained experimentally in the case of a collision between two athletes in the game of rugby.

MATERIAL AND METHODS

For our paper we are interested in evaluating the resultant force with which the ligaments and tendons of the glenohumeral joint act on the humeral head, pressing it into the glenoid cavity of the scapula. The glenohumeral joint is actively stabilized by the rotator cuff (supraspinatus, infraspinatus, subscapularis, and teres minor muscles) and passively by the glenohumeral ligaments and the joint capsule. The resultant force that maintains the humeral head in the glenoid cavity is the vector sum of these contributions.

The evaluation methods currently available are:

• computerized biomechanical modeling: 3D models of the joint, based on MRI or CT images, are used to simulate muscle and ligamentous forces in different positions of the arm. These can estimate the axial compressive force that maintains the humeral head in the glenoid;

• electromyography (EMG) combined with kinematic analysis: EMG measures muscle activity, and motion analysis allows estimation of the direction and intensity of forces. This method is indirect, but provides useful data on the contribution of the rotator cuff muscles;

• cadaveric studies: in research, anatomical samples are used to directly measure the tension of the ligaments and tendons in different positions of the shoulder. These provide approximate values of passive forces;

• advanced imaging (dynamic MRI, functional ultrasonography): can highlight the position of the humeral head and the degree of contact with the glenoid during movement, indirectly suggesting the effectiveness of the compressive forces.

Examples of clinical applications can be:

• in cases of glenohumeral instability, the evaluation of muscle and ligament forces is essential to decide between conservative treatment (kinesiotherapy) and surgical intervention;

• in rotator cuff arthropathy, the loss of compression force leads to humeral head migration and joint degeneration.

The exact values of the resultant force vary depending on the arm position, muscle tone and ligament integrity. Studies suggest that the rotator cuff can generate compressive forces of over 100 N in functional positions, but these figures are estimates and depend on the calculation method [6,7].

To develop a biomechanical model to evaluate this force, we will consider: the supraspinatus, infraspinatus, subscapularis muscles and the superior, middle and inferior glenohumeral ligaments. We adopt an axis system with the origin at the center of the humeral head. The x-axis will be horizontal and parallel to the sagittal plane. The y-axis will be vertical and parallel to the sagittal plane. The z-axis will be horizontal and parallel to the frontal plane (figure 1).

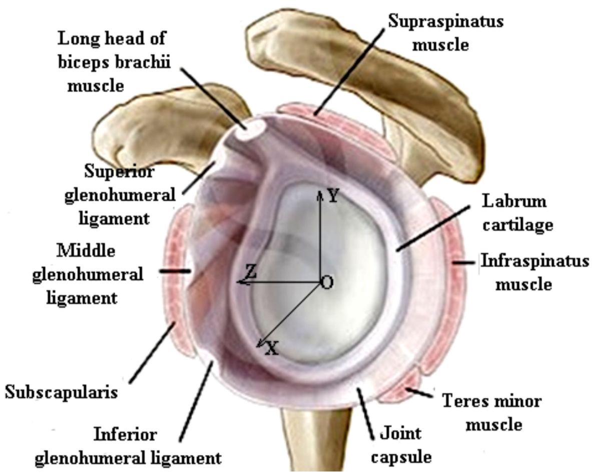

Fig. 1. The axis system with the origin in the center of the humeral head of the right shoulder (Funk, 2025)

In figures 2 and 3 see the axis system originating in the center of the humeral head of the right shoulder and the points of application of the forces exerted on the humeral head by the supraspinatus, infraspinatus, subscapularis muscles and the superior, middle and inferior glenohumeral (GH) ligaments.

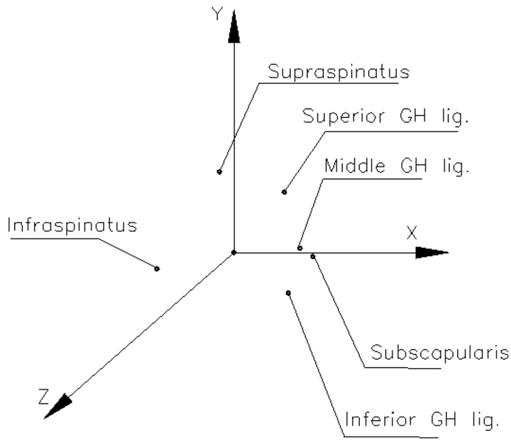

Fig. 2. The axis system with the origin in the center of the humeral head of the right shoulder and the points of application of forces on the humeral head (xy plane view) (original drawing by the authors)

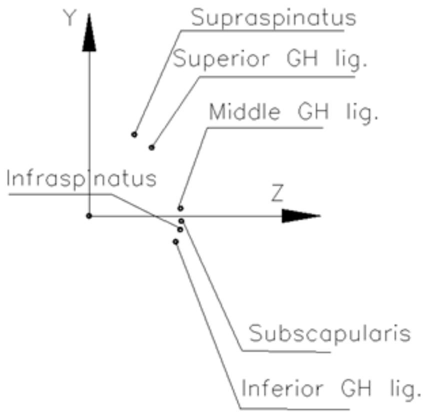

Fig. 3. The axis system with the origin in the center of the humeral head of the right shoulder and the points of application of forces on the humeral head (view in the yz plane) (original drawing by the authors)

The diameter of the humeral head in the adult male is on average between 43 and 52 mm. This value can vary slightly depending on height, constitution and ethnicity. In men, the size is generally larger than in women, due to the more developed bone and muscle mass. We will consider an average value of 49 mm.

In table 1 are presented average estimated values of the x, y and z coordinates for the application points shown in figures 2 and 3.

Table 1. The values of the geometric coordinates of the insertion points

Element (muscle or ligament)	X (mm)	Y (mm)	Z (mm)
Supraspinatus muscle	-4.37	24.33	11.64
Infraspinatus muscle	-23.15	-4.9	23.4
Subscapularis muscle	23.7	-1.21	23.5
Superior glenohumeral ligament	15.09	18.23	16.1
Middle glenohumeral ligament	19.86	1.28	23.4
Inferior glenohumeral ligament	16.29	-12.2	22.27

In table 2 are presented estimates of the angles with the x, y and z axes and of the unitary components of the forces acting on the humeral head (McMahon, 1995).

Table 2. Angles and the unitary forces

The angle and unit force of the element (muscle or ligament)	Angle with the X-axis; Unit component on the X-axis	Angle with the Y-axis; Unit component on the Y-axis	Angle with the Z-axis; Unit component on the Z-axis
Supraspinatus muscle	115°; −0.4226	71.4°; 0.3198	147°; −0.8480
Infraspinatus muscle	145°; −0.8192	60.3°; 0.4965	106.4°; −0.2870
Subscapularis muscle	80°; 0.1736	80°; 0.1736	165.8°; −0.9695
Superior glenohumeral ligament	168°; −0.9744	100°; −0.1736	98.2°; −0.1428
Middle glenohumeral ligament	110°; −0.3420	65°; 0.4226	146.7°; −0.8394
Inferior glenohumeral ligament	130°; −0.6428	45°; 0.7071	107.1°; – 0.2946

The main geometric characteristics, such as length, width, thickness, physiological section and dynamic characteristic, theoretical maximum force, for the muscles and ligaments of the glenohumeral joint and for a man with a height of 1.70 [m] and a weight between 65 and 70 [kg], are presented in table 3, with observations regarding the movement on which they intervene (Howell, 1988; Warner, 1992; Lee, 2000).

Table 3. Geometric characteristics and dynamic

Element (muscle or ligament)	Length [cm]	Width [cm]	Thickness [cm]	Physiological section area [cm²]	Theoretical maximum force [N]	Observations
Suprasinatus muscle	10-12	2-3	1-1.5	3-4	250-300	Raise the arm to abduction (the first 15-20°)
Infraspinatus muscle	11-13	3-4	1.5-2	4-5	350-450	External rotation, posterior stabilizer
Terres minor muscle	7-9	2-3	1-1.2	2-3	200-250	External rotation, stabilizer
Subscapularis muscle	13-15	4-6	2-3	6-8	500-700	Internal rotation, anterior stabilizer
Deltoideus muscle	15-18	6-8	2-3	12-15	1000-1200	The main muscle of abduction
Teres major muscle	12-14	3-5	2	6-7	450-550	Adduction and internal rotation
Superior glenohumeral ligament	25-35	4-6	1-2	-	150-200	Limits anterior translation to small abductions
Middle glenohumeral ligament	30-40	5-8	2-3	-	200-300	Anterior stability between 45-60° abduction
Inferior glenohumeral ligament	35-50	8-10	3-4	-	350-500	Main anterior stabilizer at high abduction
Coracohumeral ligament	30-40	7-10	2-3	-	400-450	Supports the joint capsule and limits external rotation

If we consider the angular positions with the reference axis system and the geometric position of the insertion points of muscles or ligaments, the projections of the traction forces are determined and presented in table 4. The unit components on the x, y and z axes in table 2 multiplied by the absolute values of the forces in the tendons and ligaments will give the components of the forces on the x, y and z axes. The resultant force acting on the humeral head will be equal to the resultant of these components.

In table 4 are presented the maximum, extreme values of the forces in the tendons and ligaments and the components on the x, y and z axes. The last row of table 4 contains the resultants on the x, y and z axes and the resultant force acting on the humeral head.

Table 4. The components of the forces along the three reference axes

Force from muscle or ligament	Maximum (extreme) [N]	Component on X [N]	Component on Y [N]	Component on Z [N]
Supraspinatus	300	−0.4226 x 300≈-127	0.3198 x 300≈96	−0.8480 x 300≈-254
Infraspinatus	450	−0.8192 x 450≈-369	0.4965 x 450≈223	−0.2870 x 450≈-129
Subscapularis	700	0.1736 x 700≈122	0.1736 x 700≈122	−0.9695 x 700≈-679
Superior gleno-humeral ligament	200	−0.9744x 200≈-195	−0.1736 x 200≈-35	−0.1428x 200≈-29
Middle glenohumeral ligament	300	−0.3420 x 300≈-103	0.4226 x 300≈127	−0.8394 x 300≈-252
Inferior gleno-humeral ligament	500	−0.6428 x 500≈-321	0.7071 x 500≈354	− 0.2946 x 500≈-147
Resultant	2450	-993	887	-1490

RESULTS

The collision of two bodies occurs in a very short time and the impact force is very high and, for this reason, in sports the effect can cause muscle and ligament trauma. The more physically prepared the athlete is and the more sports experience he has, the more he can avoid injuries. From a theoretical point of view, the collision phenomenon starts from the notions of impulse, percussion and the law of conservation of momentum (Brach, 1993; Budescu, 2008). Thus, percussion is defined for a body with the expression:

$$\mathrm{P} = \mathrm{m} \cdot (v_2 - v_1)$$

where: m – body weight [kg],
v_1, v_2 – the velocities of the body immediately before the collision and, respectively, immediately after the collision.

The multiplication between the mass of the body and its velocity represents the momentum of the body and the average percussion force is calculated with the mathematical relationship:

$$\overline{\mathrm{F}}_{\mathrm{m}} = \frac{\mathrm{m} \cdot \left(\overline{\mathrm{v}}_2 - \overline{\mathrm{v}}_1\right)}{\mathrm{t}_2 - \mathrm{t}_1}$$

where: $\Delta t = t_2 - t_1$ represents the time interval in which the collision occurs.

The experiment consisted of video recording an official rugby match, in the national rugby championship, and selecting images in which collisions occurred between two athletes (figure 4).

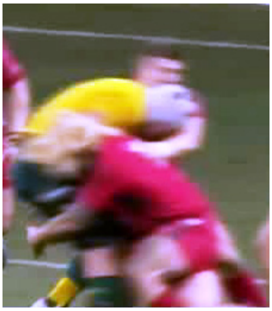

Fig. 4. The collision of two rugby players (Milburn, 2014)

After the match, the athletes of the two teams who appeared in the video recordings were identified and their weights, in kg, were determined. The video recording was then fragmented into “jpeg” images, obtaining 33 pictures for each second of recording. The measurements of the linear distances covered by the athletes, between two successive images, initially determined in "pixels" were then converted into "meters" using a calibration performed at the beginning of the video recording. Dividing the distance covered between two images by the time interval of 0.030303 seconds, the linear velocity immediately before and after the collision is obtained. In table 5 are presented the obtained values of the distances measured in pixels and meters of the athletes involved in the collision, their masses, the linear velocities then determined by calculation and average percussion force, for all seven valid video sequences captured by our recording device placed in a fixed position at the edge of the rugby field.

Table 5. Values determined based on experiment

Video sequence	Rugby player	m [kg]	Distance [pixeli]	Distance [m]	Velocity [m/s]	Average percussion force
1	1	87	74	0.2442	8.058	23134.541
	2	92	63	0.2079	6.860	20826.980
2	1	81	75	0.2475	8.167	21830.412
	2	78	91	0.3003	9.909	25505.791
3	1	75	93	0.3069	10.127	25064.350
	2	80	81	0.2673	8.8209	23287.199
4	1	87	80	0.2640	8.712	25012.177
	2	80	77	0.2541	8.382	22128.502
5	1	74	98	0.3234	10.672	26061.050
	2	95	63	0.2079	6.860	21506.121
6	1	80	78	0.2574	8.494	22424.182
	2	75	84	0.2772	9.147	22638.847
7	1	84	71	0.2343	7.731	21430.353
	2	78	83	0.2739	9.038	23263.835

In table 5, the average impact force, determined in [N], expresses the intensity exerted between the two bodies during their collision and depends on the variation of the momentum immediately before and after the impact and on the duration of contact between the bodies. There is a linear relationship between the average impact force and mechanical work, through the deformation of the body during the collision. The smaller the deformation, the greater the impact force and the kinetic energy of the body increases.

DISCUSSION

Just as in engineering, where there is a ratio between the overload supported by a shock absorber and the nominal load for which that shock absorber was designed, in the case of the joints of the human locomotor system there is such a ratio, between 2 and 10 times greater than body weight. The factors that determine the overload ratio are the following factors: the type of activity (static, dynamic, impact), the angles of the joint, the muscle tone and the time of application (Goetti, 2021; Viehöfer, 2016, Apreleva, 2000).

From table 4, if the maximum resultant is multiplied by the coefficient 10, a value of the overload force is obtained, a value that can be compared with the values of the average percussion force from table 5. In the graph in figure 5, the forces are represented, in the first column the value of the theoretical maximum force and in the following columns the values of the average percussion forces.

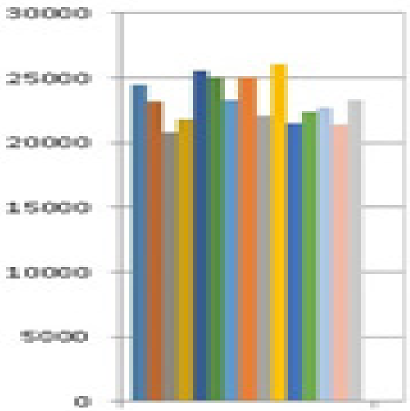

Fig. 5. Maximum theoretical force (first column) and average percussion forces (original graphic by the authors)

Due to the fact that in the calculation of the maximum theoretical force, not all muscles that contribute to the stabilization of the glenohumeral joint were taken into account, it results that the values of the average percussion force, for the analyzed experimental case, do not actually exceed the theoretical value of the joint connection force.

In the case of an athlete who has already had a shoulder dislocation, it may be recommended, in order to prevent the dislocation from recurring, to use a shoulder orthosis for preventive purposes (figure 6).

Fig. 6. Shoulder orthesis
(British Association of Prosthetists and Orthotists, 2023)

Such an orthosis ensures good stability thanks to the three adjustable straps and offers good comfort due to the material's property of regulating temperature at the shoulder level.

CONCLUSIONS

In performance sports, athletes endure mechanical joint overloads due to physical training and experience in dealing with a collision, so joint trauma is often avoided.

The presented biomechanical study offers the possibility of theoretically determining joint connection forces, using the resistive properties of the muscle and ligament tissues that stabilize the analyzed joint.

The experimental research presented in this paper can be extended to other team sports, such as basketball, handball, hockey, etc.

AUTHOR CONTRIBUTIONS

Author 1, author 2, and author 3 contributed to the design of the research, to the analysis of the results and to the writing of the manuscript. All authors have read and agreed to the published version of the manuscript.

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