Validation of OpenSim and finite element models of the human shoulder

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INTRODUCTION: Due to the complexity of studying human joint biomechanics in-vivo, non-invasive tools such as computational modelling have been developed to assess joint biomechanics. The focus of our research was to use computer modelling to study the shoulder joint, one of the most unconstrained joints of the human body. The shoulder joint is often associated with a wide range of clinical challenges, with one of the most debilitating being acromial stress fracture, a known complication of reverse shoulder arthroplasty. The two computational models used to investigate shoulder biomechanics were musculoskeletal (MSK) modelling and finite element modelling (FEM). The focus of this research was not only to develop these computational models, but to validate them using data from our in-house advanced cadaveric shoulder simulator [1]. Few previous computational models have included this validation step. In this abstract, preliminary results for a healthy shoulder FEM are presented.

METHODS: OpenSim [2] was used to create the MSK shoulder model and Abaqus (Dassault Systèmes, 2021) was used for the FEM. Both models were specimen-specific, and their bone geometries were generated from the CT scans of a cadaveric shoulder specimen. The MSK model developed was a six degree of freedom (DoF) glenohumeral joint actuated by eight Hill type muscles (Millard 2012 Equilibrium) {Anterior, Lateral, and Posterior Deltoids (AD, LD, PD respectively), Subscapularis Superior (SBS) and Inferior (SBI), Supraspinatus (SSP), Infraspinatus (ISP), and Teres Minor (TM)}. Muscle and joint reaction forces were predicted using the Concurrent Optimization of Muscles Activation and Kinematics (COMAK) algorithm [3] for one range of motion (ROM) {Abduction within scapular plane: 15° to 45° glenohumeral abduction, -30° rotation}. The validation of the MSK model was performed by comparing the kinematics and muscle forces to those generated by our cadaveric simulator for the same cadaveric specimen. To compare results between the MSK model and the cadaveric simulator, Pearson correlation coefficients were calculated, Pearson's $r > 0.5$ was considered as a good correlation. A FEM of the scapula and clavicle bones was also created from the cadaveric specimen. The mechanical properties were mapped from the CT Hounsfield units using an Abaqus plug-in, Bonemappy (https://github.com/mhogg/bonemapy.git). Loading conditions for the FEM replicated the cadaveric simulator forces of the AD, LD and PD and the joint reaction force. Two glenohumeral abduction angles were compared {15° and 45° glenohumeral abduction}. The FEM was validated by comparing the predicted bone strain at four locations along the acromion and scapular spine, as per Kerrigan et al's study [4], to strain gauge readings generated by the cadaveric simulator.

RESULTS SECTION: The MSK results are presented in Table 1. Regarding the most dominant muscle groups for the ROM tested, the results showed a good correlation (Person's $r>0.5$). For one of the non-dominant muscle groups (PD), the results showed weak correlation with a Pearson correlation coefficient less than 0.5.
Regarding the FEM results presented in Table 2, the maximum principal strains from the FEM demonstrate similar trends to the cadaveric simulator with a corresponding increase in strain across all four gauge locations as abduction angle increased. The percentage error in principal strain between the FEM and the cadaveric simulator was less than 15% for three of the four test locations with the exception of Levy zone 3A which showed an 81 % error.

DISCUSSION: A good agreement in muscle forces has been shown between the MSK model and the cadaveric simulator for the dominant muscle groups. The low correlation found for the PD muscle may be explained by its low muscle activation during abduction in both the MSK and cadaveric simulations and by the different strategies used to solve the muscle redundancy problem. The results for the FEM showed good agreement with the cadaveric simulator for three of the four zones evaluated. The discrepancy in Levy zone 3A could be due to the FEM assumption of a linear analysis. To our knowledge, this is the first shoulder FEM validation using strain gauge readings generated by a cadaveric simulator. Both MSK and FEM approaches would need to be further developed with additional samples. However, this is a promising first step to develop a double computational modelling approach allowing the assessment of potential clinical challenges such as acromion stress fracture. To our knowledge, this is the first time that a double computational modelling approach including MSK and FE analyses to investigate soft tissue injuries and surgical intervention is being developed and validated.

SIGNIFICANCE/CLINICAL RELEVANCE: Non-invasive tools, in particular computational modelling, are keys to better understand shoulder biomechanics and their validation is crucial to be able to use them for fracture prediction. This powerful tool, a double computational modelling approach, we are developing will help enhance the knowledge on the impact of soft tissue injuries and surgical intervention on the shoulder biomechanics.

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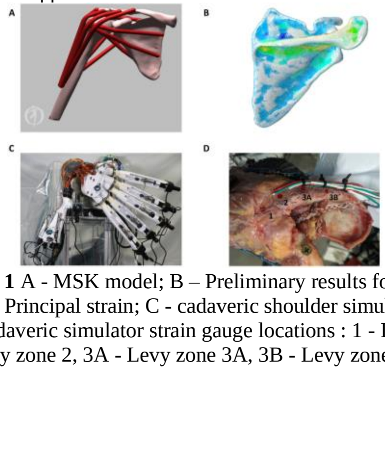

Figure 1 A - MSK model; B – Preliminary results for FEM – Max. Principal strain; C - cadaveric shoulder simulator; D - Cadaveric simulator strain gauge locations : 1 - Levy zone 1, 2 - Levy zone 2, 3A - Levy zone 3A, 3B - Levy zone 3B [4]

Table 1 *MSK Results – Pearson's Correlation Coefficient*

	Pearson's Correlation Coefficient	
	Pearson's r < 0.5	**Pearson's r > 0.5**
Muscle Groups	PD	AD, LD, SBS, SBI, SSP, ISP, TM

Table 2 *Preliminary FEM Results - Maximum Principal Strain Comparison*

	Max. Principal Strain Increase (%)	
Gauge location	**Cadaveric simulator**	**FEM**
Levy zone 1	347	326
Levy zone 2	202	232
Levy zone 3A	134	242
Levy zone 3B	160	163