Patient-specific estimation of trapeziometacarpal joint passive stiffness reveals a potential effect of hypermobility on osteoarthritis development

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Patient-specific estimation of trapeziometacarpal joint passive stiffness reveals a potential effect of hypermobility on osteoarthritis development

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Abstract

Even if osteoarthritis is multifactorial, hypermobility of the trapeziometacarpal joint is commonly considered as one of the potential risk factors since it could affect the joint mechanical loadings. Nevertheless, the results are still controversial by the lack of quantitative validation. The objective of this study was to evaluate the effect of joint laxity on trapeziometacarpal cartilage mechanical loadings. A patient-specific finite element model of the trapeziometacarpal joint passive stiffness was developed using segmentation from medical images. The joint passive stiffness was modeled by linear springs placed all around the joint. The linear spring stiffness was determined by using an optimization process to fit force-displacement data measured during laxity tests performed on eight healthy volunteers. The estimated passive stiffness parameters were then included in a full thumb finite element simulation of a pinch grip task driven by muscle forces to evaluate the effect on trapeziometacarpal loading. Correlation between the passive stiffness and the trapezium cartilage mechanical loadings in terms of joint contact pressures and maximum shear strain was analyzed. A significant negative correlation was found between the trapeziometacarpal joint passive stiffness and the trapezium cartilage contact pressure during the pinch grip task simulated. These results thus suggest that hypermobility of trapeziometacarpal joint could impact the trapezium cartilage contact pressure. Given that the risk of osteoarthritis development is associated with the intensity of cartilage loadings, our study argues in favor of the idea of an amplified risk with hypermobility.

Keywords: Trapeziometacarpal joint, Stiffness, Biomechanical modeling, Cartilage, Contact pressures, Maximum shear strain, Osteoarthritis
1. Introduction

The trapeziometacarpal (TMC) joint is the principal hand joint affected by osteoarthritis (OA) (Cvijetić et al., 2004). OA at TMC joint leads to functional capacity reduction which could be an important problem for daily life (Miura et al., 2004; Moriatis Wolf et al., 2014). Although multifactorial, the causes of this pathology are mostly linked to the mechanical loadings (principal stress, maximum shear strain, deviatoric strain, fluid velocity) endured by cartilage tissues (Buckwalter et al., 2013; Eskelinen et al., 2019; Hashimoto et al., 2009; Mononen et al., 2018; Orozco et al., 2018; Saarakkala et al., 2010; Saarakkala & Julkunen, 2010; Turunen et al., 2013; Yin & Xia, 2014). Because these internal mechanical loadings are difficult to measure, they are commonly estimated by using biomechanical models. Biomechanical models consist in representing the mechanical behavior of the anatomical structures. By solving the mechanical equations from measured or hypothesized functional boundaries, those models estimate the internal mechanical loading of anatomical structures depending on the task. Multi-body rigid (MBR) models are used to estimate the muscle forces and joint reaction forces involved in some movements such as the pinch or the power grip tasks (Barry et al., 2018; Goislard de Monsabert et al., 2014). To estimate stresses and strains in anatomic structures, the most advanced finite element (FE) models are based on CT-scan or MRI data, enabling both geometrical and material properties to be precisely defined. Muscle forces assessed by an MBR model are then included as boundary conditions for FE model of osteoarticular structures to estimate the specific mechanical loading on articular cartilage (Dong et al., 2023; Faudot et al., 2020). This modeling approach provided new insight into OA mechanical risk factors such as the influence of TMC bone morphology or grip types (Valerio et al., 2023). Among the major OA risk factors, previous studies had shown that hypermobility could also be a risk factor for TMC OA development (Jonsson et al., 1996; Jónsson et al., 2009). However, those conclusions are controversial since some other studies had shown no differences in terms of TMC joint hypermobility between healthy and OA patients (Halilaj et al., 2015). This potential risk factor thus needs to be clarified for better understanding and preventing TMC OA. TMC joint stiffness involved a complex mechanical structure of soft tissue materials, including the ligaments, the muscles, and the joint capsule (Ladd et al., 2012; Norose et al., 2022). Previous studies on ligaments and the global stiffness
characterization had shown an important interindividual variability in the TMC joint stiffness (D’Agostino et al., 2014; Domalain et al., 2010). To clarify the potential effect of hypermobility and to correctly estimate the patient-specific OA risk factor, it appears necessary to consider the specific TMC joint stiffness and its consequences on TMC mechanics. When considering FE models, the joint passive stiffness is classically modeled by taking average values of ligament stiffness from the literature, measured on cadaveric specimens (Bettinger et al., 2000; D’Agostino et al., 2014). Nevertheless, this approach is not able to clarify the contribution of individual hypermobility as a potential risk factor for OA development. Including the patient-specific stiffness in models represents a great challenge because it requires mechanical testing to determine the mechanical properties of the tissues involved in the passive stiffness (Rusli et al., 2021). Previous studies on the knee joint had used an interesting approach to estimate the material properties of ligaments based on in vivo laxity test (Ewing et al., 2016; Kang et al., 2016; Westover et al., 2016). The authors performed several laxity tests during which they recorded the force applied and the bone displacement. They identified the mechanical properties of ligaments using FE models of the knee and an optimization procedure to match experimental force-displacement data.

Consequently, to investigate the problem of joint stiffness as an OA risk, the first objective of our present study was to develop a patient-specific estimation of the TMC joint stiffness based on in vivo experimental data, inspired by methodologies performed on the knee. The second objective was to investigate the effect of joint stiffness on cartilage mechanical loadings, by using the patient-specific estimation developed in the first objective. Experimental data were based on laxity tests where force and displacement data were measured on flexion, extension, abduction, and adduction passive movements of the TMC joint, performed on healthy volunteer participants. A computational model of the TMC joint including bones and cartilage was created, based on CT-scan. Springs all around the joint were created to model the global joint stiffness. A Levenberg-Marquardt optimization method was performed to optimize the springs stiffness of the model to fit experimental data extracted from the laxity tests. Finally, patient-specific models of TMC joint stiffness were used to simulate a pinch grip task and to observe whether, as we hypothesized, TMC joint stiffness is negatively correlated with mechanical loadings, represented by principal stress and maximum shear strain in this study.
2. Methods

2.1 Stiffness experimental evaluation

Eight healthy participants (4 men, 4 women) without previous injury in the hand, aged 20 to 43 years, with first metacarpal length from 4.5 to 5.5 cm, and who signed an informed consent were included in this study. They were seated in front of a table and asked to place their right hand on the table in a neutral position (Fig.1). Their hands and forearm were placed on a custom splint to be fixed during the laxity tests. A custom experimental device was created to perform the laxity tests in passive flexion (Fig.1a), extension (Fig.1b), adduction (Fig.1c), and abduction (Fig.1d) of TMC joint. This experimental device was used to apply a force on the dorsal, volar, radial, and ulnar edge of the metacarpal bone head location identified on the participants’ skin. This experimental device was quite similar to the device used in a previous study of our group (Domalain et al., 2010) and allowed the experimenter to manually apply the load on the thumb along a specific axis. For each test, five loading cycles were performed with 2 trials separated by 2 minutes. During each laxity test, participants were asked to not resist actively the force applied. EMG data were recorded on the flexor pollicis longus (FPL) and the extensor pollicis longus (EPL) muscle to ensure that muscle activity remained negligible. A force sensor (Nano-25, ATI Industrial Automation, Garner, NC) was embedded in the custom experimental device to measure the force applied to the metacarpal head. Thumb movement during laxity tests was measured by a six-camera system (MX T40, Vicon, Oxford, UK). Three markers were placed on the first metacarpal bone dorsal side and three markers were placed on the hand dorsal side to measure the metacarpal bone position relative to the trapezium bone position (Fig.1). This protocol was approved by the local ethics committee.

Local coordinate systems were constructed from the three markers on the metacarpal and the hand plane. Orientation of the local coordinate system of the trapezium bone relative to the hand plane was estimated using a rotation matrix with constant angle values (Cooney et al., 1981). Orientation of the local coordinate system on the first metacarpal bone was then calculated relative to the coordinates system of the trapezium bone according to the previous study of (Cheze et al., 2009). In this study, Cheze et al., 2009 consider the metacarpal bone movement according to the following Euler sequence: flexion-
extension, axial rotation, and abduction-adduction. We used the anatomical landmarks from Cheze et al., 2009 to create the coordinate system on metacarpal and trapezium bones to measure the metacarpal bone displacement both experimentally and in the FE simulations. The forces measured in these laxity tests were used as boundary conditions of the FE model, presented in the next part. The displacements measured experimentally were used as targets to adjust the TMC joint stiffness in the FE model optimization.

Fig. 1 The experimental device used to measure force-displacement data of the TMC joint during passive movement of flexion (A), extension (B), adduction (C), and abduction (D). This experimental device was designed through the inspiration of the previous study of (Domalain et al., 2010).
2.2 FE model optimization to identify the subject-specific TMC joint stiffness

A 3D CAD model of the trapezium and the two first metacarpal bones in the neutral position was created with CT images (slice thickness: 0.625 mm; pixel size: 0.372 mm; resolution: 512 px × 512 px) and segmentation using Mimics (Research 22.0; Materialise, Belgium). The CT images were obtained from a previous study of our group (Valerio et al., 2023). The bone surfaces were meshed with triangles (average edge size of 1.3 mm) on 3-Matic (Research 14.0; Materialise, Belgium). The bone surface meshes were converted into volume meshes with tetrahedral elements by using FEBioStudio (Maas et al., 2012). Cartilage was extracted from subchondral bone as prismatic elements with an average thickness of 0.7 mm (Dourthe et al., 2019). Bones were modeled as rigid bodies. Cartilage was modeled as neo-Hookean hyperelastic (E = 10 MPa, ν = 0.4) (Dong et al., 2023; Kempson, 1972; Schneider et al., 2017). Contact between the trapezium and metacarpal cartilages was modeled with a sliding elastic contact implemented with a penalty-type method (Dong et al., 2023; Maas et al., 2012; Zimmerman & Ateshian, 2018). The stiffness was modeled with nine linear springs located all around the joint to represent stiffness induced by ligaments and the other tissues. The springs used were modeled using a phenomenological approach to represent the global TMC joint stiffness that potentially results from all the surrounding tissues and not only from the ligaments of the joint. The location of each spring is visible in Table 1. The force measured in the stiffness experimental evaluation (see previous part) was applied as boundary condition on the corresponding metacarpal head side to simulate the four laxity tests. The trapezium and second metacarpal bones were fully constrained with zero degrees of freedom. An optimization method was performed to adjust the stiffness of the springs to fit the displacement data measured in the experiment. Optimization was performed by the Levenberg-Marquardt method. A global objective function \( f_{\text{global}} \) was defined to minimize the square difference between the metacarpal bone displacement in all the simulations and all the experiments, by the summation of the objective function in each simulation \( f_{\text{flex}}, f_{\text{ext}}, f_{\text{add}} \) and \( f_{\text{abd}} \) as follows:

\[
\begin{align*}
  f_{\text{flex}} &= \sum_n \sqrt{(\theta(n)_{\text{flex,exp}} - \theta(n)_{\text{flex,sim}})^2} \quad (1) \\
  f_{\text{ext}} &= \sum_n \sqrt{(\theta(n)_{\text{ext,exp}} - \theta(n)_{\text{ext,sim}})^2} \quad (2)
\end{align*}
\]
\[ f_{\text{add}} = \sum_n \sqrt{\left( \theta(n)_{\text{add,exp}} - \theta(n)_{\text{add,sim}} \right)^2} \] (3)

\[ f_{\text{abd}} = \sum_n \sqrt{\left( \theta(n)_{\text{abd,exp}} - \theta(n)_{\text{abd,sim}} \right)^2} \] (4)

\[ f_{\text{global}} = f_{\text{flex}} + f_{\text{ext}} + f_{\text{add}} + f_{\text{abd}} \] (5)

\( \theta(n) \) is the metacarpal bone rotation angle at each time sample \( n \) of the corresponding laxity test (the metacarpal bone flexion angle for the flexion laxity test, and the other corresponding angle for the other tests). Different initial values of linear spring stiffness were tested to find the best initial value to get the best curve fitting. A custom Python script was developed to implement the Levenberg-Marquardt algorithm and to facilitate interaction between experimental data and the FEBio model. The finite element model was run in FEBio (Maas et al., 2012). The optimization process is summarized in Fig.2. Stiffness of each participant was then used to simulate the effect of stiffness on TMC joint mechanical loadings during a pinch grip task (see next part).

**Fig.2** The optimization process used in this study to optimize the passive stiffness of the TMC joint.

This process was performed to create a patient-specific estimation of TMC joint passive stiffness for each participant.
To investigate the effect of the TMC joint stiffness on cartilage mechanical loadings, a pinch grip task was simulated. An entire thumb with the distal and the proximal phalanx was modeled from CT images from the same patient as the optimized model but realized in a pinch grip task (Fig.3). The same process of meshing was performed to model distal and proximal phalanx bones and cartilage. Linear springs were modeled in the metacarpophalangeal and the interphalangeal joint. Their stiffness was set at 100,000 N.mm\(^{-1}\) to ensure model stability. Thumb distal phalanx was restricted to move only in her anatomical axis and degrees of freedom of the trapezium and second metacarpal bones were fully constrained. We simulated a pinch grip task with an external force of 60 N, modeling by taking the kinematic and force data, recording during this task, as input of an MBR model to estimate the muscle forces (Goislard de Monsabert et al., 2014). These muscle forces were applied as boundary conditions on the thumb FE model by identifying muscle attachment on bones and muscle force directions (Chao et al., 1989). This task was simulated with the eight springs stiffness configurations of the eight participants (obtained by the experimental evaluation of TMC joint stiffness through the optimization process), to investigate the effect of stiffness on the cartilage mechanical loadings. These simulations were performed to see if the global TMC joint stiffness is negatively correlated with the mechanical loadings in terms of joint contact pressure and maximum shear strain. Contact pressure \(p\) is defined as the average of the principal stresses \(\sigma_{11}, \sigma_{22}, \sigma_{33}\), and maximum shear strain \(\varepsilon_{\text{shr}}\) is defined as the difference between the maximum and the minimum principal strain \(\varepsilon_{\text{max}}, \varepsilon_{\text{min}}\) as follow:

\[
p = \frac{-(\sigma_{11}+\sigma_{22}+\sigma_{33})}{3} \quad (6)
\]

\[
\varepsilon_{\text{shr}} = \varepsilon_{\text{max}} - \varepsilon_{\text{min}} \quad (7)
\]

These two values were calculated at each node, then averaged in the joint contact area on the trapezium cartilage surface. The global TMC joint stiffness is defined as the sum of the stiffness of each spring for the corresponding participant. These simulations were also run in FEBio (Maas et al., 2012).
Fig.3 The finite element model of the thumb joint developed to study the TMC joint stiffness effect on joint mechanical loadings during a pinch grip task. The linear springs placed in the joints are represented in green and the muscle forces applied are represented by red arrows. Black triangles represent the degrees of freedom removed for the distal phalanx, the trapezium, and the second metacarpal. This model was run in FEBio.

2.4 Statistical analysis

Linear regression analyses were performed, considering TMC joint contact pressure and maximum shear strain as response variables and participant stiffness as explanatory variable. A Pearson test was used to evaluate the significance of the correlation. The significance threshold was set at $p < 0.05$.

3. Results

3.1 Spring stiffness determined by the laxity test

The mean force applied on the metacarpal head during the laxity tests were $5.2 \pm 1.0$ N, $7.8 \pm 3.3$ N, $10.9 \pm 1.2$ N, and $8.0 \pm 2.3$ N for the flexion, abduction, adduction, and extension tests respectively. The flexion, abduction, adduction, and extension rotation angle reached $0.10 \pm 0.03$ rad, $0.20 \pm 0.09$ rad, $0.12 \pm 0.05$ rad, and $0.24 \pm 0.06$ rad at the end of the respective tests. An example of the force-displacement data obtained during experiments and of the fitted curve obtained by the optimization is visible in Figure 4. The normalized root mean square error (RMSE) between experimental and simulated
force-displacement data averaged 16.9 %. The sum of each linear spring stiffness (which defined the
global TMC joint stiffness) averaged 132.1 ± 22.9 N.mm\(^{-1}\) (range: 105.7 – 182.6 N.mm\(^{-1}\)). The stiffness
for one spring only ranges from 4.9 to 24.7 N.mm\(^{-1}\). The stiffness of each spring obtained by the
optimization and the RMSE for all the participants is presented in detail in Table 1.

**Fig.4** Comparison between experimental (Raw data) and simulation (Fitted curve) data. This figure
shows the force-displacement data of the abduction laxity test from participant P3. The fitted curve is
obtained by optimizing the stiffness of each spring in the model to match the raw data angular
displacement with the Levenberg-Marquardt method. On this curve, the normalized RMSE is 10.9 \%.
Table 1 Results of the optimization performed to optimize each spring stiffness to fit force-displacement data of each participant’s laxity test. This table shows the stiffness and the location of each spring, the sum of each spring which represents the global stiffness of the TMC joint (for the correlation test), and the normalized angle RMSE.

<table>
<thead>
<tr>
<th>Spring Location</th>
<th>Participant</th>
<th>P1</th>
<th>P2</th>
<th>P3</th>
<th>P4</th>
<th>P5</th>
<th>P6</th>
<th>P7</th>
<th>P8</th>
</tr>
</thead>
<tbody>
<tr>
<td>Spring 1</td>
<td>Dorso-Central</td>
<td>5.1</td>
<td>16.8</td>
<td>24.7</td>
<td>24.7</td>
<td>5.4</td>
<td>12.8</td>
<td>18.8</td>
<td>24.5</td>
</tr>
<tr>
<td>Spring 2</td>
<td>Dorso-Ulnar</td>
<td>23.7</td>
<td>5.5</td>
<td>11.9</td>
<td>23.1</td>
<td>9.5</td>
<td>24.7</td>
<td>24.5</td>
<td>24.2</td>
</tr>
<tr>
<td>Spring 3</td>
<td>Ulno-Volar</td>
<td>5.1</td>
<td>24.4</td>
<td>24.6</td>
<td>5.4</td>
<td>21.0</td>
<td>21.8</td>
<td>10.6</td>
<td>5.0</td>
</tr>
<tr>
<td>Spring 4</td>
<td>Centro-Volar</td>
<td>15.2</td>
<td>11.0</td>
<td>8.3</td>
<td>19.5</td>
<td>12.3</td>
<td>4.9</td>
<td>24.4</td>
<td>18.7</td>
</tr>
<tr>
<td>Spring 5</td>
<td>Centro-Radial</td>
<td>5.1</td>
<td>20.0</td>
<td>4.9</td>
<td>22.2</td>
<td>7.0</td>
<td>4.9</td>
<td>6.8</td>
<td>13.0</td>
</tr>
<tr>
<td>Spring 6</td>
<td>Dorso-Radial</td>
<td>20.3</td>
<td>5.0</td>
<td>5.0</td>
<td>19.8</td>
<td>14.2</td>
<td>4.9</td>
<td>24.5</td>
<td>12.3</td>
</tr>
<tr>
<td>Spring 7</td>
<td>Centro-Volar</td>
<td>17.0</td>
<td>11.6</td>
<td>24.5</td>
<td>11.6</td>
<td>4.9</td>
<td>7.0</td>
<td>23.5</td>
<td>4.9</td>
</tr>
<tr>
<td>Spring 8</td>
<td>Dorso-Radial</td>
<td>5.0</td>
<td>5.6</td>
<td>5.2</td>
<td>16.6</td>
<td>23.3</td>
<td>4.9</td>
<td>24.7</td>
<td>4.9</td>
</tr>
<tr>
<td>Spring 9</td>
<td>Radio-Volar</td>
<td>9.1</td>
<td>21.8</td>
<td>24.7</td>
<td>6.1</td>
<td>24.3</td>
<td>24.7</td>
<td>24.6</td>
<td>24.7</td>
</tr>
<tr>
<td>Sum</td>
<td></td>
<td>105.7</td>
<td>121.6</td>
<td>133.9</td>
<td>149.0</td>
<td>122.0</td>
<td>110.9</td>
<td>182.6</td>
<td>132.4</td>
</tr>
</tbody>
</table>

Note. The normalized angle RMSE corresponds to the RMSE divided by the maximum angle measured during each laxity test.

3.2 TMC joint contact pressure and maximum shear strain during pinch grip task

TMC joint contact pressures and maximum shear strain distribution is presented in Figure 5. Among the eight simulations tested with the stiffness of the eight participants, seven simulations only terminated successfully. Results of the TMC joint pressure for the participant where the simulation was unsuccessful (the participant P6) are not presented.

The TMC joint contact pressure and maximum shear strain averaged 9.3 ± 0.6 MPa and 0.29 ±0.05 respectively. The TMC joint contact pressure for the participants P1, P2, P3, P4, P5, P7, and P8 are respectively 10.0 MPa, 9.8 MPa, 8.9 MPa, 9.8 MPa, 9.5 MPa, 8.1 MPa, and 9.4 MPa. The maximum shear strain is respectively 0.32, 0.31, 0.22, 0.34, 0.33, 0.22, and 0.30. The maximal difference in joint contact pressure is observed between participants P1 and P7 with a difference of 23.4 %. The maximal difference in maximum shear strain is observed between participants P4 and P3/P7 with a difference of 54.5 %.
Fig. 5 Distribution plot of the TMC joint contact pressures (A) and maximum shear strain (B) in the trapezium cartilage for each participant. Plot of the participant P6 is not presented due to the unsuccessful simulation. This figure shows the contact pressure and the maximum shear strain at the end of the simulation.

### 3.3 Correlation between TMC joint stiffness and TMC joint mechanical loadings

Pearson test reveals a significant correlation between TMC joint stiffness and TMC joint contact pressures ($r^2 = 0.65; p < 0.05$). The linear regression plots of the relationship between the TMC joint stiffness and the TMC joint contact pressures and maximum shear strain are presented in Figure 6.
Fig. 6 Linear regression between TMC joint contact pressure, maximum shear strain on cartilage, and TMC joint stiffness of each participant with the successful simulation. The averaged TMC joint contact pressure and maximum shear strain for each participant are represented by blue points and the regression lines by red lines.

4. Discussion

The objectives of this study were to propose a method to estimate patient-specific TMC joint passive stiffness and to clarify the effect of TMC joint stiffness on joint mechanical loadings and its potential implication in OA risks. To achieve this goal, laxity tests were performed on eight healthy humans to estimate patient-specific joint stiffness by inverse finite element analysis. Then, a finite element model of the entire thumb in a pinch grip task was used to investigate the effect of stiffness on cartilage contact pressures and maximum shear strain. A linear regression was performed to analyze the correlation between TMC joint stiffness and cartilage contact pressures and maximum shear strain.

The global TMC joint stiffness values obtained in the current study agreed with previous studies and varied between participants. Our parameter identification optimization indeed provided values that are consistent with the literature on ligament stiffness characterization (Bettinger et al., 2000; D’Agostino et al., 2014). The global stiffness as the sum of all spring stiffnesses averaged 132 N.mm\(^{-1}\) in this study.
(Table 1) against 154 N.mm\(^{-1}\) in (D’Agostino et al., 2014) and 159 N.mm\(^{-1}\) in (Bettinger et al., 2000).

Although the cited authors only considered the contribution of the main TMC ligaments in joint stiffness without considering other anatomical structures (muscles, skin, capsule), their values are higher than those computed in the current study. This comparison suggests that stiffness could be slightly underestimated by the present method. This difference could be explained by the bone displacement estimation error induced by the soft tissue artifacts, which can cause an error in TMC joint angle estimation (Kuo et al., 2003). Another explanation could be the position of the force application point on the metacarpal head in the optimization process which was visually estimated from the force sensor placement. Nonetheless, error was minimized by taking the same trained experimenter. The normalized RMSE obtained in the current study between experimental and simulation are in the same range as previous studies on the knee (Ewing et al., 2016; Kang et al., 2016; Westover et al., 2016). The maximal normalized RMSE was reported by (Westover et al., 2016) with 94.29% between experimental and simulation which explains possible important errors with this kind of approach. Unlike the current study, these previous works only modeled the ligaments and thus neglected other anatomical structures participating in the global stiffness. The better agreement between experiments and simulation obtained in this study is thus probably related to the choice of a phenomenological approach, i.e., representing the global stiffness by linear springs all around the joint. This choice was made after running the first simulations (not reported here) attempting to represent the anatomical location of TMC ligaments but could not find reasonable agreement. Several studies showed that TMC ligamentous structures are complex with large inter-individual variations which prevent the representation by average anatomy (Bettinger et al., 1999; D’Agostino et al., 2014; Nanno et al., 2006). Because of these inter-individual variations and probably because other structures such as muscle or skin are involved in TMC stiffness, the approach consisting in representing anatomical paths of ligaments appears not suited for TMC.

Despite a lack of significant correlation, the FE pinch grip simulations suggest that maximum shear strain could be related to TMC passive stiffness. The strain values indeed varied importantly between participants with up to 54.5% inter-individual differences. The maximum cartilage shear strain averages 0.29 for all the participants, which is higher than the average values reported by a previous study on the knee (Chan et al., 2016). This previous study reported femoral and tibial cartilage maximum shear strain.
slightly below 0.1. This important difference could be explained by the mechanical properties used to model the cartilage in this study. NeoHookean hyperelastic model could represent correctly the mechanical response of cartilage at the macroscopic level and justify why it is commonly used for this kind of approach (Dong et al., 2023; Faudot et al., 2020b; Schneider et al., 2017). However, the fluid part of the cartilage is omitted with this approach and could influence the stress-strain relationship even so. A more complex cartilage model should be used in further studies to integrate the tissue complexity and improve the model’s accuracy. This more complex approach will also provide some other interesting variables like cartilage fluid velocity, which is one of the best variables to understand cartilage degeneration according to previous literature (Orozco et al., 2018). Beyond modeling questions, our study already suggests a relatively important inter-individual variation in shear strain on TMC cartilage that needs to be studied in light of a potential risk factor of OA.

The significant correlation found between the TMC joint stiffness and the joint contact pressures (Fig.6) confirmed our hypothesis. This correlation could be explained by the metacarpal bone movement increase and the consequences on the cartilage contact. Our results indicate also a possible overpressure of 23.4 % (the maximal difference observed in the results) between the higher and the lower stiffness. These findings indicate a significant effect of passive stiffness on TMC contact pressures and thus the necessity to individualize it in the FE models. A previous study on the knee joint had shown the same conclusions with a sensitivity analysis performed on a knee FE model (Rooks et al., 2022). Considering cartilage response to high mechanical loadings, this overpressure of 23.4 % could in the long term increase the risk of OA development (Buckwalter et al., 2013; Eskelinen et al., 2019; Hashimoto et al., 2009; Mononen et al., 2018; Orozco et al., 2018; Saarakkala et al., 2010; Saarakkala & Julkunen, 2010; Turunen et al., 2013; Yin & Xia, 2014). Some limitations should be considered. Stiffness is not the unique factor that could be correlated with joint contact pressure. An important interindividual variability was also reported on TMC bony morphology (Rusli & Kedgley, 2020) and could also influence joint contact pressure (Schneider et al., 2017). A recent study of our group had shown that trapezium dorso-vo lar curvature is correlated with joint contact pressure during pinch grip tasks (Valerio et al., 2023). In this previous study, joint contact pressure ranges from 6.8 to 15.8 MPa with various morphology against 8.1 to 10.0 MPa with various stiffness in this study. Further studies should test the
combined effect of morphology and stiffness to see how they interact and to see if the variations could
be more important with the interaction. This study focuses on the stiffness effect on one pinch grip task
in one joint position and previous literature had shown that the task can influence pressure distribution
in the TMC joint (Schneider et al., 2017). Other TMC joint positions should be tested in the simulations
to see the interaction between the stiffness and the posture.

Despite these limitations, this study provided an innovative method to create patient-specific estimations
of TMC joint stiffness and highlight the necessity to individualize this parameter in the computational
models to ensure good accuracy. Our results indicate a correlation between stiffness and joint contact
pressure and can suggest an effect of hypermobility on OA development. Further studies must
investigate the stiffness effect in different configurations, with more samples and in interaction with the
other parameters to confirm these findings.

**Author contributions**

All authors have made significant contributions to the study design. Thomas Valerio, Jean-Louis Milan,
Benjamin Goislard de Monsabert and Laurent Vigouroux contributed data analysis and interpretation.
Development of the model, experimental setup, and drafting of the paper were performed by Thomas
Valerio. All authors contributed critical revision of the paper and approved the final version of the
manuscript to be published.

**Conflict of interest**

The authors declare no conflict of interest and have no relevant financial or non-financial interests to
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