The effect of index finger distal interphalangeal joint arthrodesis on muscle forces and adjacent joint contact pressures

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Abstract

Distal interphalangeal joint arthrodesis is a frequent surgical operation performed to treat severe arthritis. Nevertheless, the angle selected when fusing the joint is arbitrarily chosen without any quantified data concerning its mechanical effects, thus preventing the optimal choice for the patient. In the current study, we realized an experiment and developed a numerical model to investigate the effect of fusion angle on the biomechanics of adjacent non-operated joints. Six participants performed a pinch grip task while arthrodesis was simulated with a metal splint. Kinematic and force data were recorded during this task and used in a biomechanical model to estimate contact pressures in adjacent joints. The biomechanical model involved combining a multibody system and a finite element method. Results showed that the angle of any distal interphalangeal joint arthrodesis influences index finger kinematics and maximal grip force in several participants. For one participant, in the arthrodesis simulation, we observed an increase of 1.9 MPa in the proximal interphalangeal joint contact pressure. Our results provide quantified information about the biomechanical consequences of this surgical operation and its potential long-term effects.

Keywords Osteoarthritis · Arthrodesis · Distal interphalangeal joint · Musculoskeletal modelling · Finite element modelling

1 Introduction

The distal interphalangeal (DIP) joint of the index finger is the area most frequently affected by osteoarthritis in the hand [9]. Palliative treatments like splints, pain killers, and physiotherapy are sometimes insufficient to decrease the pain and recourse to surgical solutions is thus required [19].

At the DIP joint, arthrodesis is the most recommended operation [5]. It consists in fusing the distal and medial index phalanges. This fusion is performed by inserting an implant in the bone diaphysis. The most commonly used implants are K-wires, compression screws, and intramedullary implants like the X-Fuse implant (X-Fuse®, Stryker, USA). The X-Fuse implant differs from the others by its geometry and mechanical properties which offer good small-bone fusion [2, 10, 27]. In particular, it offers several fusion angle options corresponding to 0, 15, and 25° of the DIP joint flexion. Nevertheless, the lack of scientific data prevents any quantified and scientifically validated choice for fusion angles. As a result, the selection is often based on subjective criteria. Zero degrees is commonly chosen for aesthetic reasons [10], whereas 15° or 25° are recommended for manual workers.

At this stage, there is no consensus among biomechanical engineers or clinicians regarding the optimal fusion angle for DIP joint arthrodesis to maximize functional recovery [5, 10, 22]. Most biomechanical studies have focused on proximal interphalangeal (PIP) joint arthrodesis [4, 11, 15] with only one study concerning the DIP joint [22]. These authors have shown that a slightly flexed angle, around 20°, could be more

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effective in improving grip force and manual dexterity than an angle of zero degrees (0°).

However, such conclusions were made by only using grip force measurements and manual dexterity without considering the internal biomechanics of the corresponding finger. Domalain et al. [11], found that for the PIP joint arthrodesis, the fusion angle can have an influence on the kinematics of other joints. This point is of importance since modifying joint angles may influence the internal mechanics of the osteoarticular system. The effect of arthrodesis on grip force and joint position influences muscle forces, as their production and transmission capacity also depend on joint position, through the force–length relationship and moment arm values [17, 26]. Ultimately, such changes at the muscle level could also influence joint contact pressures [14]. The DIP fusion angle could thus affect the biomechanics of adjacent non-operated joints altering finger kinematics, muscle forces, and joint contact pressures. Higher pressures in those joints could increase cartilage degeneration and lead to the early development of osteoarthritis [6, 13].

Given these considerations, it appears crucial to help surgeons in the choice of the DIP fusion angle to study how the index DIP joint arthrodesis (IDIPJA) angle affects the biomechanics of the entire finger. The aim of this study is to investigate the biomechanical consequences of IDIPJA and the associated fusion angle on grip capacities. First, we analyzed the direct impact of the fusion angle on grip force and index finger kinematics. Secondly, we investigated how muscle function was affected by different fusion angles. Finally, the impact at the osteoarticular level was quantified by studying how the DIP fusion angle influences the joint contact pressure in adjacent non-operated joints. To answer this objective, healthy participants performed a pinch grip task, while IDIPJA was simulated with splints. Joint kinematics and grip forces were recorded, and a hybrid biomechanical model of the index was then used to evaluate both muscle forces and joint contact pressures. According to previous literature, we hypothesized that high fusion angles lead to higher grip force, lesser index posture modification, lower muscle forces and lower contact pressure in adjacent joints.

2 Methods

2.1 Participants and protocol

Six healthy right-handed participants (5 male and 1 female) were recruited for this study (age: 30.3 ± 8.4 years; height: 173.2 ± 8.6 cm; weight: 67.8 ± 9.1 kg; hand length: 18.8 ± 1.3 cm; hand width: 8.7 ± 0.5 cm). Each participant was free of any upper right extremity disorder and signed an informed consent. This protocol was approved by the Aix-Marseille University Ethics Committee.

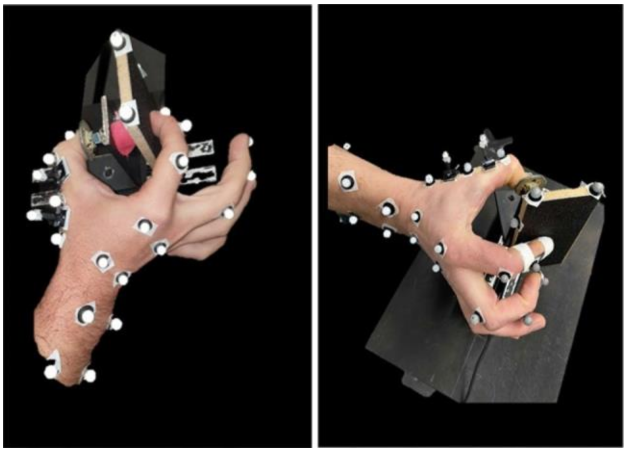
Each participant was seated in front of a table and performed a pinch grip task. The pinch grip task consisted in grasping a 5.5-cm-long object between the fingertips of the right hand thumb and index (Fig. 1a). Participants were asked to seize and raise the object at a comfortable height and to exert a maximal force on it for 5 s. IDIPJA was simulated by positioning a splint guiding both the dorsal and volar parts of the distal and medial phalanges (Fig. 1b). Three fusion angles were simulated in line with those proposed for X-Fuse implant: 0° (A0), 15° (A15), and 25° (A25). A control condition (C) without splint was also performed, giving a total of four conditions. Two trials were performed for each condition and were separated by a 2 min resting period to prevent fatigue entering into the equation. Grip force and kinematic data were simultaneously recorded and synchronized for each trial. Only the data corresponding to the highest grip force peak trial were used for the analysis.

2.2 Force measurement

A six-axis force sensor (Nano-25, ATI Industrial Automation, Garner, NC) was embedded in the object to measure the force applied by the index finger and thumb (Fig. 1). The six signals were recorded at 2000 Hz, filtered (Butterworth, 4th order, cut-off frequency: 20 Hz), re-sampled at 100 Hz (kinematic sampling), and averaged over a 750-ms window centered on the grip force peak. The three force components were fed into the musculoskeletal model and applied at the middle of the index distal phalanx.

2.3 Kinematic analysis

The 3D position of hand and forearm segments was recorded at 100 Hz using a seven-camera Qualysis Oqus optoelectronic system (Qualysis; Göteberg, Sweden) with twenty-three 6-mm diameter spherical markers (Fig. 1). The marker set involved tracking the direct kinematics of dorsal bony landmarks [23] and comprised five markers for the index and middle fingers, six for the thumb, and four for the wrist. T-clusters were used to track the first metacarpal and the thumb proximal phalanx [8]. Three other markers were placed on the force sensor to determine the measured force's orientation relative to the finger segments. Each marker coordinate was averaged over the 750-ms window centered on the grip force peak. Joint angles were calculated from the orientation of the distal segment relative to the proximal segment for each joint. The Euler/Cardan angles method has been used to extract joint angles. The sequence used for this method was flexion/abduction/pronation as this was the sequence used by Chao in the musculoskeletal model [7]. Joint angles of the ring and little fingers were considered as the same as the middle finger to simplify the kinematic data acquisition protocol.



(a) (b)

Fig. 1 Experimental device used for kinematic and force data acquisition. This figure shows the force sensor and the 23 markers used for two of the four experimental conditions: **a** control condition and **b** simulated IDIPJA with DIP flexion angle of 0°

2.4 Musculoskeletal multibody rigid model

The musculoskeletal (MSK) model was developed in the paper by Goislard De Monsabert et al. [16, 17]. Briefly, this model consisted in estimating the tendon and muscle forces by solving mechanical equilibrium equations of the all hand joints using numerical optimization, to account for muscular redundancy, based on muscle stress criteria. The static moments of all degrees of freedom were equilibrated using the following equation:

$$[R] \times \{t\} + m_L + m_F = \{0\}$$

where $[R]$ represents the muscle moment arms, $\{t\}$ represents the muscle tension, $\{m_L\}$ represents the passive moments, and $\{m_F\}$ represents the moments of external forces.

The optimization method used to solve the previous undetermined equation system was based on the following cost function:

$$\min \sum_m \left(\frac{(t_m)_s}{PCSA_m}\right)^4$$

where $(t_m)_s$ is the muscle tension for s solution and $PCSA_m$ the physiological cross sectional area of m muscle.

This model includes 42 muscles and 23 degrees of freedom. It considers each bone segment as a rigid body and the joints as mechanical linkages without friction. The model estimated output muscle forces with kinematic and force data as the input. Only the forces of the six index finger muscles were analyzed in this study, namely flexor digitorum profundus (FDP), flexor digitorum superficialis (FDS), long extensor (LE) which is the resulting muscle of index extensor and common finger extensor, lumbrical (LU), ulnar interosseous (UI), and radial interosseous (RI). The muscle forces of these muscles were estimated for each participant in each condition. For the condition of simulated arthrodesis, the DIP joint flexion degree of freedom was removed from the mechanical equations solved in the optimization procedure, assuming the distal and medial phalanges as a single rigid segment. Muscle force outputs from this model were then used as inputs in the finite element model.

2.5 Finite element modelling

The finite element (FE) model of the index finger was developed in a previous study by our group [14]. Basically, this mechanical model reproduces the three-dimensional structure of the osteoarticular system including bones, cartilage, ligaments, tendons, and pulleys (Fig. 2). Bone geometry was reconstructed by segmentation and all the other tissues were reconstructed with anatomical features identified on bone. Linear elastic isotropic material was used for modelling cortical bones (E = 18 GPa, ν = 0.2), cancellous bones (E = 300 MPa, ν = 0.25), tendons (E = 3 GPa, ν = 0.3), and pulleys (E = 1 GPa, ν = 0.3). Cartilage was modelled as Neo-Hookean hyperelastic material (C10 = 0.34 MPa, D1 = 2.20 MPa^{-1}). Ligaments were modelled with nonlinear elastic springs (with stiffness from 40 to 150 N/mm). Muscle forces estimated by the musculoskeletal rigid multibody model were applied at the associated tendons' extremities and joint contact pressures of the PIP and MCP joints were estimated by the finite element method. The latter was driven by the Abaqus/Explicit analyzer (SIMULIA, Dassault System®). All the finite element model parameters, except muscle forces, such as mesh size, interaction, and boundary conditions, were the same as in the previous study [14]. For the simulated arthrodesis condition, the DIP joint angle was modified by rotating the distal phalanx around the DIP rotation axis (Fig. 2). To simulate arthrodesis bone fusion, a Tie constraint was applied, preventing any relative motion of bone joint surfaces of the DIP. PIP and MCP joint contact pressures were estimated for each participant and each condition by average contact pressures of each node of the joint contact area.

2.6 Analysis

Dependent variables for this study were maximal grip force, PIP. and MCP joint flexion angle, muscle force ratios for FDP, FDS, RI, LE, UI, and LU and joint contact pressure for the PIP

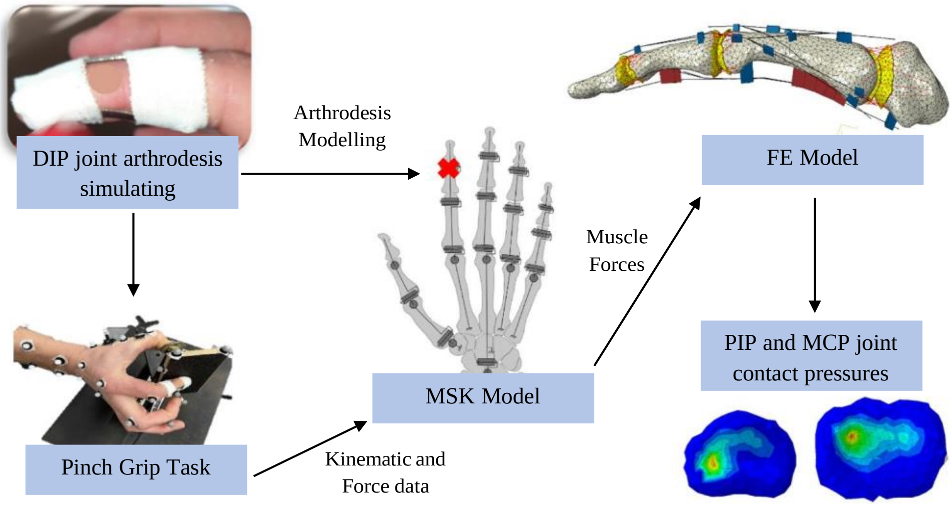


Fig. 2 Schematic of the methodology used in this study. Index DIP joint fusion was simulated with a metallic splint. A pinch grip task was performed to produce kinematic and force data. These data were inserted into a musculoskeletal (MSK) model to estimate muscle forces which were used in a finite element (FE) model to estimate PIP and MCP joint contact pressures. MSK and FE models were adapted to simulate DIP joint arthrodesis

and MCP joints. Muscle force ratios correspond to muscle forces normalized by the grip force experimentally measured for each participant in each condition. These variables were individually analyzed for the six participants and the four experimental conditions. Due to the small number of participants and trials, only result comparisons between the conditions were made and no statistical analysis was performed.

3 Results

3.1 Grip force and joint angles during pinch grip task

The grip forces observed for the six participants in the four conditions are presented in Fig. 3. Grip force averaged 49.5 ± 11.8 N, 48.6 ± 10.9 N, 44.9 ± 6.3 N, and 44.3 ± 7.2 N for the conditions C, A0, A15, and A25 respectively. Changes in grip force were observed for all participants but in different amounts. Fusion angle influenced grip force levels achieved by participants P1, P4, and P6, by up to 17.3 N, 17.4 N, and 8.7 N respectively. No major variations between the four conditions were observed for participants P2, P3, and P5, with differences less than 4.0 N.

The joint flexion angles in the four conditions are presented in Fig. 4. The PIP joint flexion angle averaged 37.7 ± 15.5°, 38.1 ± 9.4°, 37.9 ± 10.6°, and 34.6 ± 7.1° for the conditions C, A0, A15, and A25 respectively. Fusion angle influenced the PIP joint flexion angle for participants P4, P5, and P6 by up to 21.4°, 18.1°, and 21.8° respectively. No major differences were observed for participants P1, P2, and P3, with changes remaining below 10° of the PIP joint flexion angle across the four conditions (except for participant P2 with 11.6°). The MCP joint flexion angle averaged 32.8 ± 15.2°, 29.9 ± 14.2°, 29.1 ± 14.3°, and 28.0 ± 13.1° for the conditions C, A0, A15, and A25 respectively. A major difference in MCP joint flexion angle was observed only for the participant P1, with a maximal variation of 14.8°. No major differences were observed for participants P2, P3, P4, P5, and P6, for whom changes remained less than 10°.

3.2 Muscle force ratios

MSK modelling revealed that the main muscles involved during the pinch grip task for all conditions are FDP, FDS, RI, and LE, with the LU and UI muscles being less involved, especially LU for which muscle forces did not exceed 15 N across all participants and conditions. For this reason, only FDP, FDS, RI, and LE muscle force ratio results are presented.

The FDP, FDS, RI, and LE muscle force ratios are presented in Fig. 5. FDP muscle force ratios averaged 1.7 ± 0.8, 1.9 ± 0.9, 1.8 ± 0.8, and 1.7 ± 0.7 for the conditions C, A0, A15, and A25 respectively. The corresponding values for the other muscles were as follows FDS: 1.6 ± 0.9, 1.5 ± 1.1, 1.6 ± 0.8, and 1.8 ± 1.0, RI: 2.4 ± 0.5, 2.8 ± 0.8, 2.6 ± 0.7, and 2.7 ± 0.5, LE: 1.1 ± 0.4, 0.9 ± 0.4, 0.7 ± 0.5 and 0.8 ± 0.5.

For FDP, the bigger differences between the fusion angle conditions were observed for participants P1 and P6 with muscle force ratio increases of up to 1.1 and 1.0 respectively. Differences in muscle force ratios observed for participants P2, P3, P4, and P5 were 0.3, 0.5, 0.3, and 0.7 respectively.

For FDS, the larger differences were observed for participants P4 and P6 with muscle force ratio increases of up to 2.3 and 2.0 respectively. Differences in muscle force ratios observed for participants P1, P2, P3, and P5 were 1.3, 0.5, 0.8, and 0.7 respectively.

For RI, the only major difference was observed for participant P4 with a muscle force ratio increase of up to 1.0. Differences in muscle force ratios observed for participants P1, P2, P3, P5, and P6 were 0.4, 0.6, 0.6, 0.3, and 0.5 respectively.

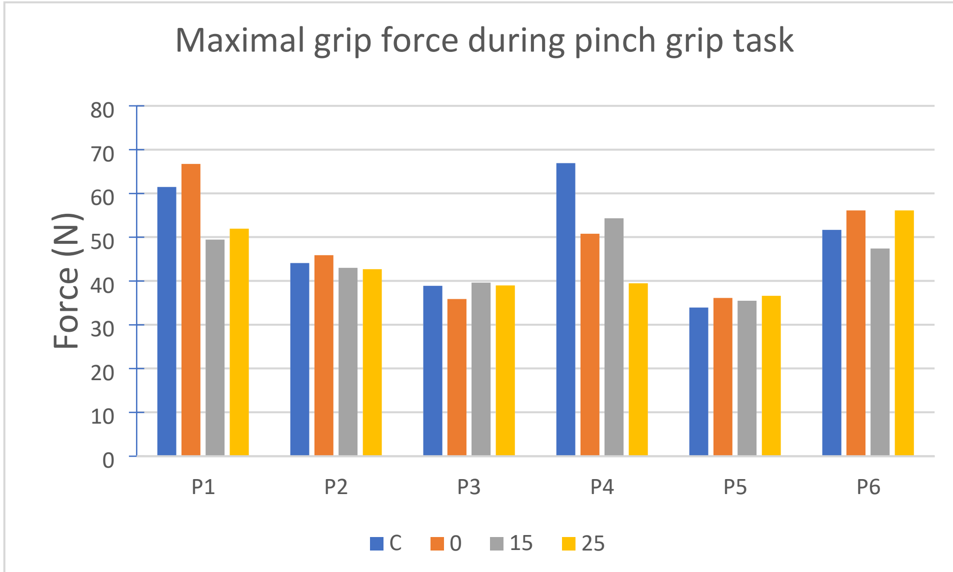


Fig. 3 Maximal grip force measured during pinch grip task in the four experimental conditions (Control, 0°, 15°, 25°) and for the six participants, from participant 1 (P1) to participant 6 (P6). This figure shows maximal values of the average grip force over a 750-ms window centered on the grip force peak during the trial with the highest grip force peak

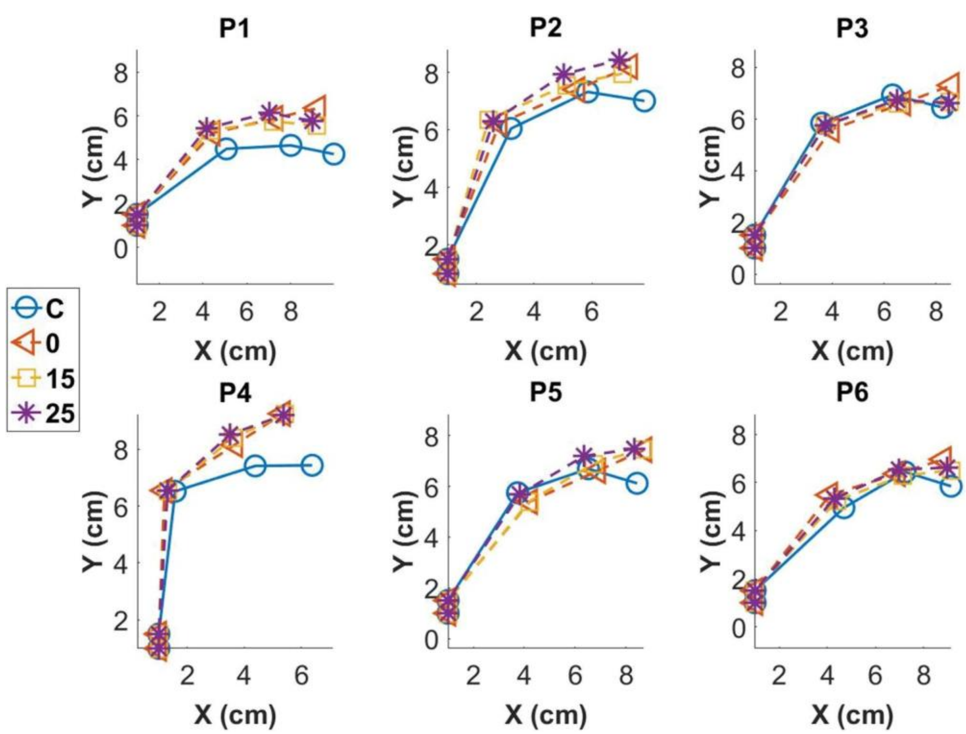


Fig. 4 Index sagittal view of the six participants (P1, P2, P3, P4, P5, P6) in the four experimental conditions (Control, 0°, 15°, 25°). The X axis represents the anteroposterior axis and the Y axis represents the dorsovolar axis. The 3 index phalanges are represented by segments. Joints and index pulp are represented by a circle for the condition C, by a triangle for the condition 0, by a square for the condition 15 and by an asterisk for the condition 25, which represents, from left to right: index metacarpal extremity, MCP joint, PIP joint, DIP joint and index pulp

For the LE muscle, the sole major difference was observed for participant P5 with a muscle force ratio increase of 1.0. Differences in muscle force ratios observed for participants P1, P2, P3, P4, and P6 were 0.7, 0.5, 0.4, 0.4, and 0.4 respectively.

3.3 Adjacent joint contact pressures

The adjacent joint contact pressures at the PIP and MCP joints, estimated by the FE model for the four conditions and the six participants, are presented in Fig. 6. PIP joint contact pressure values were 4.4 ± 0.8, 4.7 ± 1.2, 4.2 ± 1.1, and 4.2 ± 0.7 MPa for the conditions C, A0, A15, and A25 respectively. MCP joint contact pressure values were 5.2 ± 0.8, 5.5 ± 1.4, 5.0 ± 1.2, and 5.1 ± 1.1 MPa for the same conditions. For the PIP joint, the fusion angle influenced the joint contact pressure of participant P4 with the value increasing up to 1.9 MPa. Differences in joint contact pressures observed for participants P1, P2, P3, P5, and P6 were 1.2, 1.0, 0.2, 0.5, and 1.6 MPa respectively. For

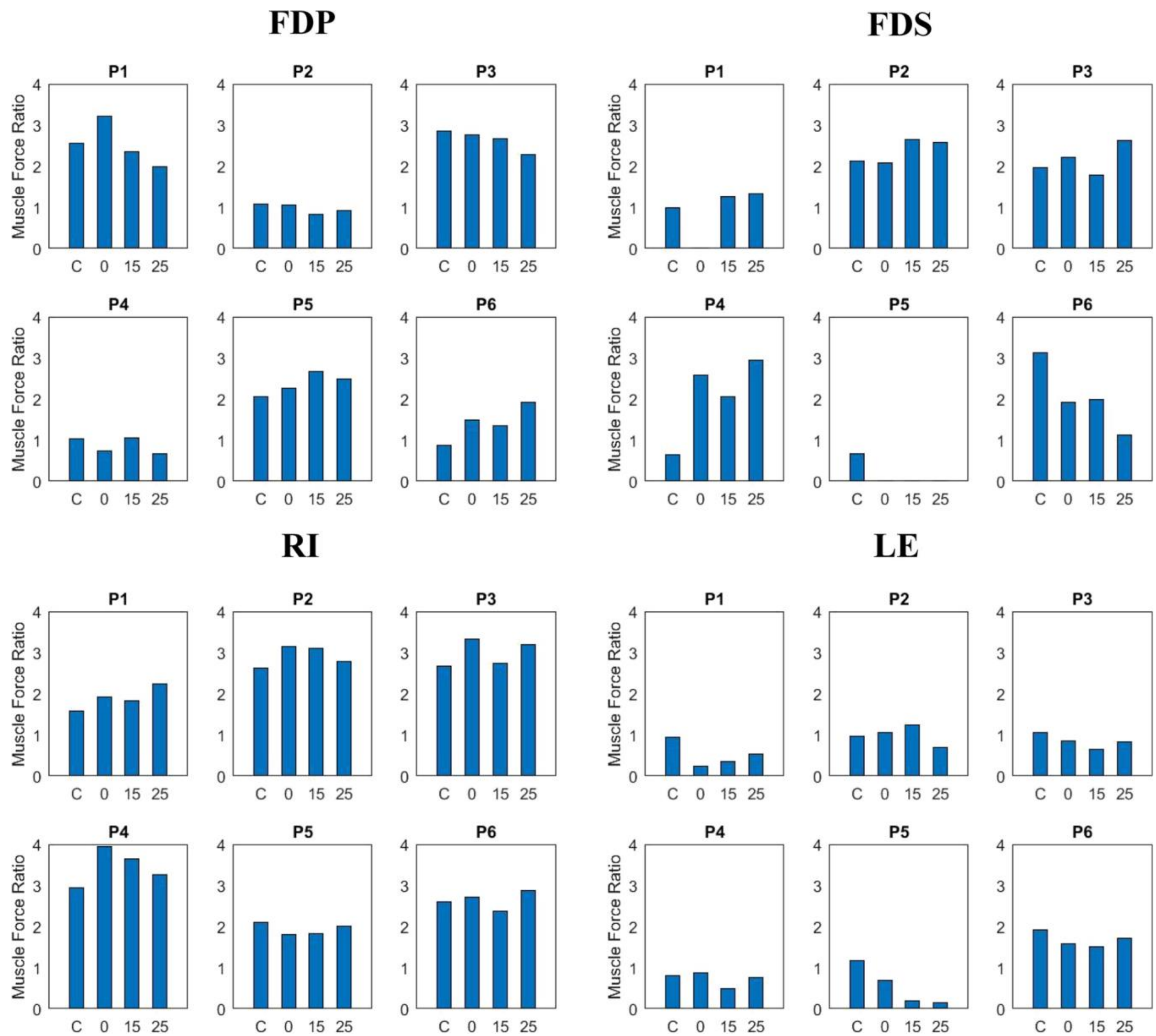


Fig. 5 Index muscle force ratio for the six participants (P1, P2, P3, P4, P5, P6) in the four experimental conditions (Control, 0°, 15°, 25°). Four of the six index muscles are presented here: FDP, FDS, RI, and LE. Each value represents muscle forces estimated by the MSK model (Goislard de Monsabert et al., 17) with kinematic and force data. Muscle force ratio represents the muscle force normalized by maximal grip force during the pinch grip task

the MCP joint, the fusion angle influenced joint contact pressure of participant P4 with the value increasing up to 1.7 MPa. Differences in joint contact pressures observed for participants P1, P2, P3, P5, and P6 were 0.8, 0.7, 1.0, 0.8, and 1.1 MPa respectively.

4 Discussion

4.1 Summary

The objectives of this study were to investigate biomechanical consequences of IDIPJA and the associated fusion angle, on grip force, finger kinematics, muscle forces, and joint contact pressures. Three types of measurement were made. Muscle forces were estimated by using a musculoskeletal multibody rigid model and their ratios were calculated. Finally, joint contact pressures were estimated by a finite element

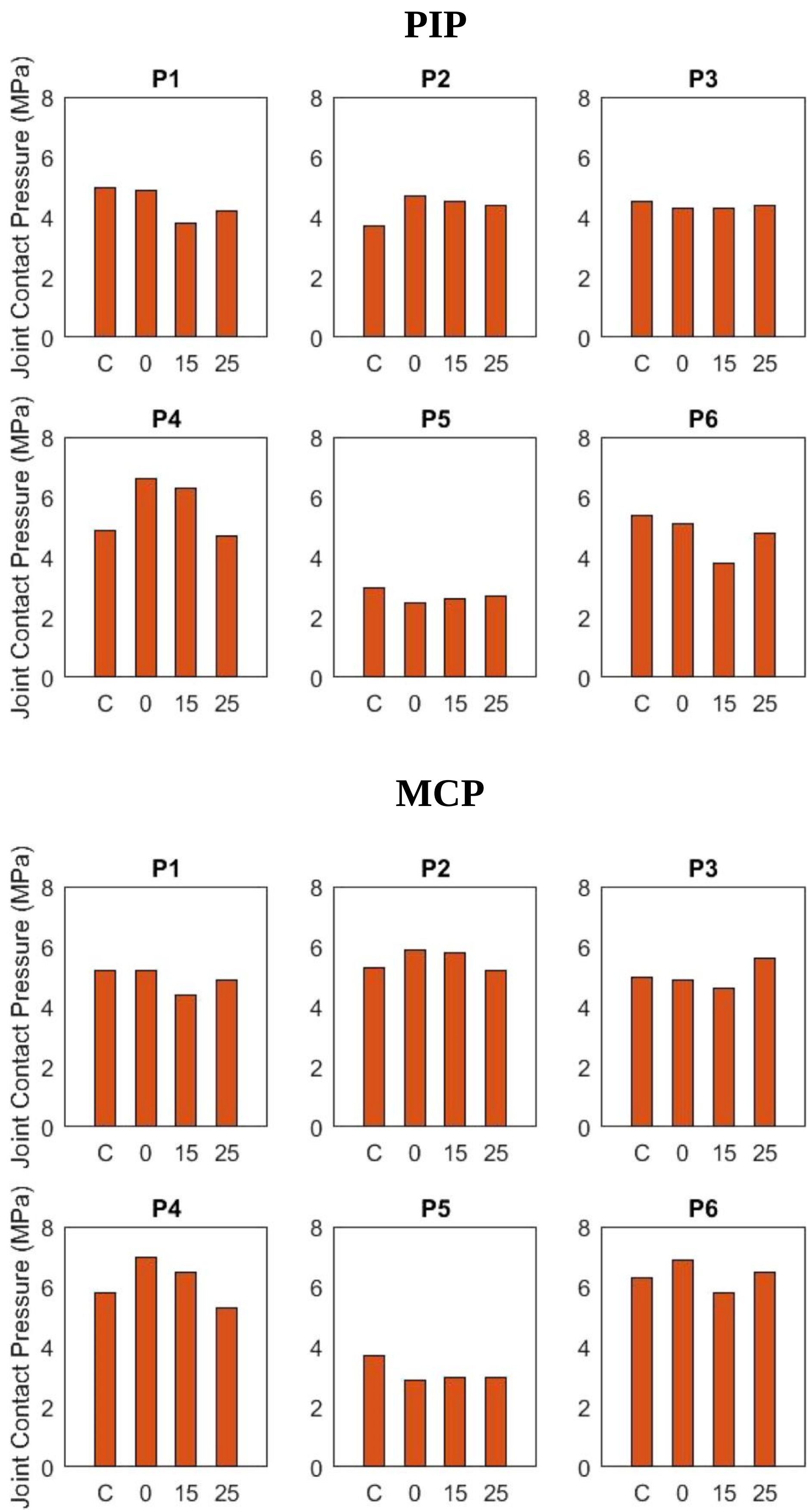


Fig. 6 PIP and MCP joint average contact pressure for the six participants (P1, P2, P3, P4, P5, P6) in the four conditions (control, 0°, 15° and 25°). These pressure values are calculated by the finite element model [14], by average node pressure values in the joint contact area

model driven by muscle forces. These three different levels of interpretation are presented in the following paragraphs.

4.2 Arthrodesis effect on grip force and finger kinematics

Maximal grip force levels were consistent with the study of Domalain et al. [12] on a similar pinch grip task, despite the large variability among the six participants. A greater sample might have led to a reduced inter-variability. Maximal grip force variations according to the four conditions differ considerably between each participant. Two groups were distinguishable with a clear influence of fusion angle for participants P1, P4, and P6 while participants P2, P3, and P5 remained stable. This difference can be explained by the fact that the DIP joint fusion affects patients differently as observed in previous studies [10]. For the three participants showing an influence of fusion angle, maximal grip force was higher for a fusion angle at 0° and higher than or equal to the one measured at 25° (Fig. 3). This force variation is not in accordance with the study of Melamed et al. which showed that maximal grip force increases with fusion angle [22]. This difference could be due to a difference in the grip task. Melamed et al., 2014 used a smaller pinch width of 3 cm against 5.5 cm in this experiment and studies have shown that this width influences index posture and maximal grip force [12]. This parameter seems to evolve differently by participant and according to the object size.

Concerning the joint angles, PIP and MCP joint angle levels are consistent with the study of Domalain et al. [12] which used the same object size. Results show important index posture adaptation for only two participants: participant P1 and participant P4 (Fig. 4). Participant P1 mainly modified the MCP joint angle with a more extended posture for the three arthrodesis conditions in comparison to the control condition. Participant P4 mainly modified the PIP joint position, also through more extended postures for the three arthrodesis conditions. This suggests that IDIPJA simulation can lead to adjacent joint posture adaptations, but these adaptations were highly dependent on the participant. It could be assumed that these different types of adaptation were partly due to the anthropometric differences between the participants. The hand length of participant P1 is 20.6 cm against just 17.7 cm for participant P4. They both are the extrema of the averaged population [18]. We can thus suppose that this brings important preferential posture differences in pinch grip tasks according to Ambike et al. [1].

4.3 Arthrodesis effect on index muscle forces

Muscle force ratio levels obtained by the musculoskeletal model were comparable with those in the study by Vigouroux et al. [28]. FDS force ratio turns out zero for the participants P1 and P5 in arthrodesis conditions. This is probably due to the distal interphalangeal removal from the musculoskeletal model which leads to a higher FDP contribution in the other joints. Consequently, FDP tendon force is enough to equilibrate the index joints, thus reducing the FDS contribution. Results revealed different variations of muscle force ratios during the four conditions depending on the participant. The largest variation in muscle force ratios across all participants and all muscles were observed for the FDS muscle of participant P4 (Fig. 5) with differences between conditions of up to 2.0. These results can be explained by the changes in index joint angles previously discussed, especially by the PIP joint flexion which is less than in the control condition. Given that, the FDS muscle moment arm in the PIP joint increases with PIP joint flexion [3], and the muscle force required to produce the same external moment is reduced. The lower influence of fusion angle on muscle force ratios observed for other participants might be due to the fact that index posture modifications across conditions were also less pronounced (Fig. 4). Our results thus suggest that arthrodesis simulation can lead to a muscle force ratio that is 4.5 times higher than without arthrodesis. Assuming IDIPJA leads to the same effect for patients, these muscle force modifications could have deleterious effects on global hand function. Such muscle force increases could increase the risk of localised fatigue, tendinitis or other muscle and tendon damage ultimately reducing the patients' functional capacities [25].

4.4 Arthrodesis effect on adjacent joint contact pressures

Joint contact pressure levels estimated by the finite element model ranged between 2 and 8 MPa which is consistent with the findings of Faudot et al. [14]. The largest effect of fusion angle on adjacent loading was observed for participant P4, especially for PIP where the difference in joint contact pressure between control condition and a fusion of 0° is 1.9 MPa (Fig. 6). This result is not surprising as this participant also showed the largest variations of finger posture and muscle force. It confirms results from previous studies showing an intricate link between abilities (applied force and finger posture), muscle forces, and joint loading [14, 17] and justifying the need for more studies to understand the outcomes of such surgical acts.

By combining all these measurements and results, this study showed that arthrodesis simulation can lead to participant specific adaptations of index finger posture, inducing an increase in muscle forces ultimately resulting in higher contact pressure on the joint's cartilage. Despite the effect of arthrodesis simulation being minimal for most of the participants, a large increase in contact pressure was observed in participant P4. From a clinical perspective, these results suggest that IDIPJA could have an impact on adjacent joint contact pressures. Considering that mechanical loading is one of the most important factors concerning osteoarthritis development [13], more studies are needed to clarify the consequences of IDIPJA on early osteoarthritis onset in the PIP and MCP joints. This aspect is important to take into account for the surgeon considering the importance of PIP and MCP for finger mobility [20].

Our results indicated that an accurate investigation should be conducted on the patient specificity to clarify the IDIPJA effect on finger biomechanics and help surgeons in their decisions, especially concerning patients' anthropometry. The consequence on mechanical loading and the pathological risk at adjacent joints could be highly amplified according to each individual. In particular, surgeons should be attentive to patients with hand lengths greater than 90% of the overall population, and patients privileging some pinch grip posture (namely with PIP and MCP extended).

Some limitations should be considered. First, we focused on IDIPJA simulation and a non-real surgical operation, whereas surgery greatly modifies patients' situation on other levels than on the anatomical and biomechanical ones [10]. For instance, grip capability could be influenced by a reduced tactile sensibility induced by the operation [24] and thus modify biomechanical consequences on the musculoskeletal system. Another limitation is the task used for this study, which is a pinch grip task with maximal force, which is not representative of everyday activities. It should also be noted that the musculoskeletal multibody rigid and finite element models used in this study are associated with inherent limits, such as the modeling hypothesis and lack of a direct validation method. Nevertheless, these approaches were currently the only means to estimate internal forces and pressures in complex musculoskeletal systems like the hand. In addition, the number of participants was low in this study thus preventing the use of any statistical tests. Finally, it should be mentioned that our study focused solely on the mechanical aspect of osteoarthritis whereas this pathology is multifactorial [21].

In spite of these limitations, this study seems to provide some understanding of biomechanical consequences of IDIPJA. Even if results do not enable clarifying the problem of fusion angle choice, the study reveals that this particular surgical operation could have detrimental consequences for patients. Further studies should focus on evaluating patients' abilities and internal biomechanics pre and post-surgical intervention to clarify the outcomes of IDIPJA and the significance of the fusion angle.

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Declarations

Conflict of interest The authors declare no competing interests.

References

1. Ambike S, Paclet F, Zatsiorsky VM, Latash ML (2014) Factors affecting grip force : anatomy, mechanics, and referent configurations. Exp Brain Res 232(4):1219–1231. https://doi.org/10.1007/s00221-014-3838-8
2. Ameline T, Bégot V, Ardouin L, Hulet C, Hanouz N (2015) Arthrodesis of thumb interphalangeal and finger distal interphalangeal joints using the intramedullary X-Fuse® implant : retrospective analysis of 38 cases. Chir Main 34(2):67–72. https://doi.org/10.1016/j.main.2015.01.002
3. An KN, Chao EY, Cooney WP, Linscheid RL (1979) Normative model of human hand for biomechanical analysis. J Biomech 12(10):775–788. https://doi.org/10.1016/0021-9290(79)90163-5
4. Arauz P, DeChello K, Dagum A, Sisto SA, Kao I (2017) Biomechanics and pinch force of the index finger under simulated proximal interphalangeal arthrodesis. J Hand Surg 42(8):658.e1-658.e7. https://doi.org/10.1016/j.jhsa.2017.04.002
5. Beldner S, Polatsch DB (2016) Arthrodesis of the metacarpophalangeal and interphalangeal joints of the hand : current concepts. J Am Acad Orthop Surg 24(5):290–297. https://doi.org/10.5435/JAAOS-D-15-00033
6. Buckwalter JA, Anderson DD, Brown TD, Tochigi Y, Martin JA (2013) The roles of mechanical stresses in the pathogenesis of osteoarthritis : implications for treatment of joint injuries. Cartilage 4(4):286–294. https://doi.org/10.1177/1947603513495889
7. Chao, E. Y. (Éd.). (1989). *Biomechanics of the hand : a basic research study*. World Scientific.
8. Cooney WP, Lucca MJ, Chao EY, Linscheid RL (1981) The kinesiology of the thumb trapeziometacarpal joint. J Bone J Surg Am 63(9):1371–1381
9. Cvijetić S, Kurtagić N, Ozegović DD (2004) Osteoarthritis of the hands in the rural population : a follow-up study. Eur J Epidemiol 19(7):687–691. https://doi.org/10.1023/b:ejep.0000036794.40723.8e
10. De Almeida YK, Athlani L, Dap F, Dautel G (2019) Distal interphalangeal joint arthrodesis using the X-Fuse® implant : a retrospective study of 54 fingers with 24 months' follow-up. Hand Surg Rehabil 38(3):186–190. https://doi.org/10.1016/j.hansur.2019.01.001
11. Domalain M, Evans PJ, Seitz WH, Li Z-M (2011) Influence of index finger proximal interphalangeal joint arthrodesis on precision pinch kinematics. J Hand Surg 36(12):1944–1949. https://doi.org/10.1016/j.jhsa.2011.09.010
12. Domalain M, Vigouroux L, Danion F, Sevrez V, Berton E (2008) Effect of object width on precision grip force and finger posture. Ergonomics 51(9):1441–1453. https://doi.org/10.1080/00140130802130225
13. Droz-Bartholet F, Verhoeven F, Prati C, Wendling D (2016) Prevention of hand osteoarthritis by hemiparesis. Arthritis & Rheumatol 68(3):647–647. https://doi.org/10.1002/art.39512
14. Faudot B, Milan J-L, Goislard de Monsabert B, Le Corroller T, Vigouroux L (2020) Estimation of joint contact pressure in the index finger using a hybrid finite element musculoskeletal approach. Comput Methods Biomech Biomed Engin 23(15):1225–1235. https://doi.org/10.1080/10255842.2020.1793965

15. Fram BR, Seigerman DA, Cross DE, Rivlin M, Lutsky K, Bateman MG, Watkins C, Beredjiklian PK (2020) The Optimal position for arthrodesis of the proximal interphalangeal joints of the border digits. J Hand Surg 45(7):656.e1-656.e8. https://doi.org/10.1016/j.jhsa.2019.11.008
16. Goislard De Monsabert B, Rossi J, Berton É, Vigouroux L (2012) Quantification of hand and forearm muscle forces during a maximal power grip task. Med Sci Sports Exerc 44(10):1906–1916. https://doi.org/10.1249/MSS.0b013e31825d9612
17. Goislard de Monsabert B, Vigouroux L, Bendahan D, Berton E (2014) Quantification of finger joint loadings using musculoskeletal modelling clarifies mechanical risk factors of hand osteoarthritis. Med Eng Phys 36(2):177–184. https://doi.org/10.1016/j.medengphy.2013.10.007
18. Guerra RS, Fonseca I, Pichel F, Restivo MT, Amaral TF (2014) Hand length as an alternative measurement of height. Eur J Clin Nutr 68(2):229–233. https://doi.org/10.1038/ejcn.2013.220
19. Kloppenburg M, Kroon FP, Blanco FJ, Doherty M, Dziedzic KS, Greibrokk E, Haugen IK, Herrero-Beaumont G, Jonsson H, Kjeken I, Maheu E, Ramonda R, Ritt MJ, Smeets W, Smolen JS, Stamm TA, Szekanecz Z, Wittoek R, Carmona L (2019) 2018 update of the EULAR recommendations for the management of hand osteoarthritis. Ann Rheum Dis 78(1):16–24. https://doi.org/10.1136/annrheumdis-2018-213826
20. Leibovic SJ (2007) Arthrodesis of the interphalangeal joints with headless compression screws. J Hand Surg 32(7):1113–1119. https://doi.org/10.1016/j.jhsa.2007.06.010
21. Marshall M, Watt FE, Vincent TL, Dziedzic K (2018) Hand osteoarthritis : Clinical phenotypes, molecular mechanisms and disease management. Nat Rev Rheumatol 14(11):641–656. https://doi.org/10.1038/s41584-018-0095-4
22. Melamed E, Polatsch DB, Beldner S, Melone CP (2014) Simulated distal interphalangeal joint fusion of the index and middle fingers in 0° and 20° of flexion : a comparison of grip strength and dexterity. J Hand Surg 39(10):1986–1991. https://doi.org/10.1016/j.jhsa.2014.06.021
23. Metcalf CD, Notley SV, Chappell PH, Burridge JH, Yule VT (2008) Validation and application of a computational model for wrist and hand movements using surface markers. IEEE Trans Biomed Eng 55(3):1199–1210. https://doi.org/10.1109/TBME.2007.908087
24. Miall RC, Rosenthal O, Ørstavik K, Cole JD, Sarlegna FR (2019) Loss of haptic feedback impairs control of hand posture : a study in chronically deafferented individuals when grasping and lifting objects. Exp Brain Res 237(9):2167–2184. https://doi.org/10.1007/s00221-019-05583-2
25. Sande LP, Cote Gil Coury HJ, Oishi J, Kumar S (2001) Effect of musculoskeletal disorders on prehension strength. Appl Ergon 32(6):609–616. https://doi.org/10.1016/S0003-6870(01)00035-7
26. Rockenfeller R, Günther M (2017) How to model a muscle's active force–length relation : a comparatiVE study. Comput Methods Appl Mech Eng 313:321–336. https://doi.org/10.1016/j.cma.2016.10.003
27. Seitz WH, Marbella ME (2013) Distal interphalangeal joint arthrodesis using nitinol intramedullary fixation implants : X-fuse implants for DIP arthrodesis. Tech Hand Up Extrem Surg 17(3):169–172. https://doi.org/10.1097/BTH.0b013e31829ba688
28. Vigouroux L, Domalain M, Berton E (2011) Effect of object width on muscle and joint forces during thumb–index finger grasping. J Appl Biomech 27(3):173–180. https://doi.org/10.1123/jab.27.3.173

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