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Force ratio in masticatory muscles after total replacement of the temporomandibular joint

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The temporomandibular (TM) joint is one of the most active joints in the human body, and any defect in this joint has a significant impact on the quality of life. The objective of this study was to analyze changes in the force ratio after TM joint replacement on contralateral TM joint loading. Implantation of an artificial TM joint often requires removal of 3 of the 4 masticatory muscles (activators). In order to perform true loading of the TM joint, loading during mastication was investigated. Input kinematic variables and mastication force were experimentally examined. The inverse dynamics approach and static optimization technique were used for solution of the redundant mechanism. Muscle forces, and reactions in the TM joint were calculated. We modified the model for several different tasks. The *m. temporalis* and *m. masseter* were removed individually and together and the forces of mastication on the TM joint were calculated for each variation. To evaluate the results, a parametric numerical FE analysis was created to compare the magnitude of the TM joint loading during the bite process for four different muscle resections. The results show an influence relative to the extent of muscle resection on contralateral TM joint loading in a total TM joint replacement. The biggest increase in the loading magnitude on the contralateral TM joint is most evident after *m. masseter* and *m. temporalis* resection. The results from all simulations support our hypothesis that the greater the extent of muscle resection the greater the magnitude of contralateral TM joint overloading.

Key words: *inverse dynamics, static optimization, temporomandibular joint, muscle forces, mastication, FE analyses*

1. Introduction

Up to 60% of the population suffers from temporomandibular (TM) disorders, with females being affected more than males. From an anatomical and mechanical point of view, the TM joint is a complicated bicondylar joint complex, and is one of the most loaded joints in the human body. Its uniqueness lies in the fact that the TM joint is actually two identical joints. With both joints occurring on the same bone, the mandible. Any movement or dysfunction of one joint also affects the joint on opposite site. Understanding temporomandibular (TM) joint reactions and musculotendon masticatory forces, as well as development of a realistic musculoskeletal mathematical model [5] of the lower jaw-skull system has been a matter of considerable interest since the 1980s [15], [14]. Data from research on this joint are important in many medical [18], [9] and scientific fields such as dentistry, maxillofacial surgery, biomechanics [12], as well as being necessary for a better understanding of bone and joint loading.

Currently, three types of total TM joint replacements are used in clinical practice. All three total replacements have a similar condylar type replacement surgical implantation method, which is, according to our analysis, essential for achieving optimal function. During implantation of the condylar part, the *m. masseter* is always resected from the lateral surface of the *ramus mandibulae*, and in most cases so is the *m. temporalis* together with *processus coronoideus mandibulae*. This method of fixing the replacement elements to the bone causes a radical change in the

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force and kinematic ratios during mandibular loading of both TM joints. Due to the resection of the *m. masseter*, the muscles on the other side must supply the force needed for mastication, bite, and overall stability of the whole system; however, this leads to an undesirable increase in loading on the contralateral TM joint.

The objective of this work was to analyse loading magnitude and distribution in the TM joint on both sides after resection of masticatory muscles during implantation of an artificial TM joint. Results were compared using parametric numerical simulations of the whole TM joint system using the Finite Element Method (FEM). When there is increased loading, we must respect these forces during dimensioning of the artificial TM joint replacement. The main goal of this study was to investigate the sensitivity of load distribution after a partial resection of masticatory muscles.

2. Materials and methods

A 3D mathematical model of the mandible and skull, including sixteen actuators for two TM joints, was first created using the MATLAB software system (MathWorks, Inc.). All simulations were created on a rigid body model of the whole TM joint system without a TM disc. Actuators included the *m. masseter* 2× (deep and surface), *m. temporalis* 2× (anterior and posterior), *m. pterygoideus medialis* 2× (anterior and posterior) and *m. pterygoideus lateralis* 2× (inferior and superior); the result was 8 activators for each (left and right) side [1]. The model consisted of the mandible; skull, muscle attachments positions, and was symmetrical around the median plane. The mandible can move with six degrees of freedom relative to the skull and the distance between TM joints was assumed to be invariable. Muscle attachments and muscle parameters such as physiological cross-sectional area, pennation angle, muscle mass; optimum muscle lengths, etc., were taken from the literature [1].

The mandible and skull 3D motion was experimentally analyzed (by three cameras and processed using APAS software (Ariel Dynamics Inc.)) during bolus processing. An experimental measurement of kinematic parameters was used to simultaneously investigate masticatory force during soft bolus mastication. The masticatory force was measured using a 3D force micro-sensor. The force sensor was implanted into the side of the specimen bolus. This method for measuring *in vivo* masticatory forces was described in [2]. A total of six equilibrium equations were derived for the mandible. These equations contained sixteen unknowns (in the form of actuators forces) and two unknown reactions in the TM joints, each in three directions. The input kinematics and masticatory force for muscle and TMJ force calculations were taken from three mastication movements.

A representation of the musculotendon complex (Hill type model) using idealized mechanical objects is expressed in equation (1). This is the basis for the physiological EMG-driven model [7], [16] which considers factors related to force-velocity f_v, force-length f_l^{act} and activation level $a(t)$ of the contractile muscle component [19], force-length relation of passive muscle component f_l^p, maximum isometric muscle force F_0^M, and pennation angle $\alpha(t)$. This model corresponds with a full Hill type musculotendon complex.

$$F = F_0^M [f_l^{act} \cdot f_v \cdot a(t) + f_l^p] \cos(\alpha(t)) . \qquad (1)$$

An EMG recording for activation signal estimation of all single and deep muscles is practically impossible. One potential method of addressing these discrepancies is to use an optimization scheme that assumes that EMGs are inherently imperfect, and the driven activation signal $a(t)$ (normalized EMG) is calculated using an optimization method. Therefore, in place of actuators forces the unknown variable now becomes activation (for each muscle). This problem was solved using the constrained static optimization technique. Optimization criteria were minimization of muscle activation, equation (2), and minimization of TM joint reactions, equation (3).

$$\mathbf{J} = \sum_{i=1}^{n} a_i^2 , \qquad (2)$$

$$\mathbf{J} = \sum_{i=1}^{n} R_i^2 . \qquad (3)$$

The function described by equation (1) was constrained so that all of the musculotendon forces were positive ($F_i \geq 0$) since muscles cannot produce compressive force. Limitation of an activation level was next inequality constraints $1 \geq a_i \geq 0$. After calculation of muscle and TMJ forces during mastication, the same forces were calculated, while leaving out the muscles that are usually removed during implantation (i.e., *m. masseter* and *m. temporalis*).

For evaluation of results obtained from simulations we used parametric model for numerical simulations using FEM, where we analyzed the influence of surgical procedures used during a total TM joint replacement on the loading of the contralateral joint. The numerical simulations were based on a simplified deformable model of the mandible, the inferior dental arch, both TM joint discs and the fossa. The geometric model was designed on the basis of CT images of a healthy patient without obvious damage to the mandible and TM joint. In the created FE simulations four simulations were modelled of the influence of the surgery while implanting the total replacement: (a) resection caput mandibulae while the *m. pterygoideus lat.* was removed, (b) resection of caput mandibulae with removing of *m. pterygoideus lat.* and *m. masseter*, (c) resection behind processus coronoideus mandibulae while *m. temporalis* and *m. pterygoideus lat.* were removed, (d) the last resection behind processus coronoideus mandibulae while *m. masseter*, *m. temporalis* and *m. pterygoideus lat.* were removed. In the FE model, the total temporomandibular joint replacement was modelled as a spherical joint. More detailed information about the parametric FE model of the whole TM joint system is presented in [4].

3. Results

One of the aims in this work was an experimental measurement of masticatory force during mastication. Masticatory force was measured using a 3D force micro-sensor [8] and was measured in only one place, the most loaded tooth (i.e., the first molar), on the side of specimen bolus, although under real situations the mandibles make contact at multiple points, see graph

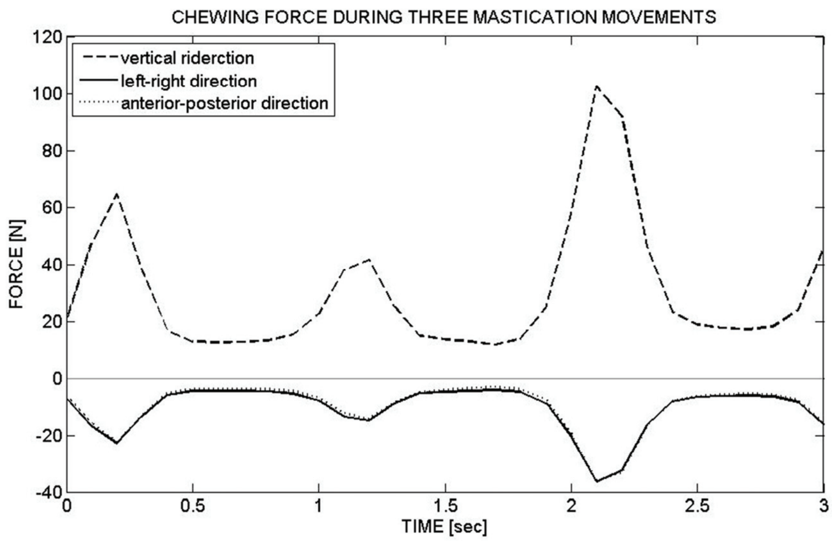

Fig. 1. Experimentally measured masticatory force on balance side during three mastication movements

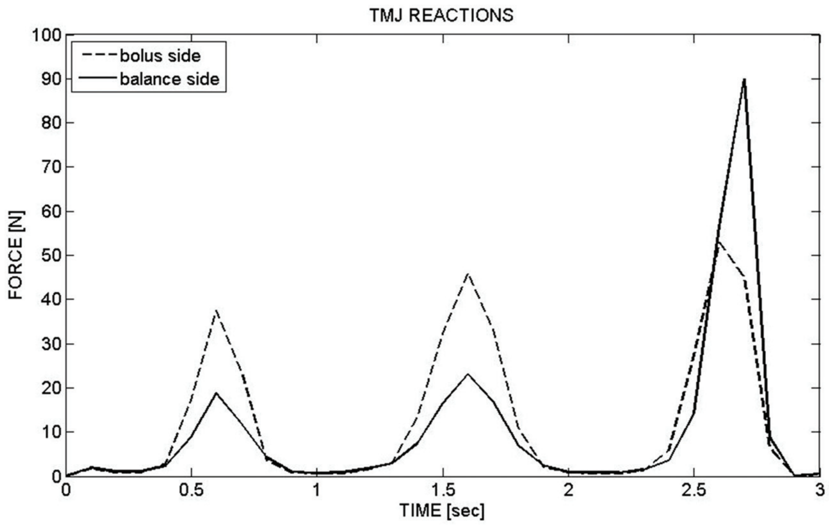

Fig. 2. TMJ reactions calculated from experimentally measured masticatory and movement data

in Fig. 1. Masticatory forces in both left-right and anterior-posterior directions are not insignificant. Magnitudes of forces are about 30% in the left-right and anterior-posterior directions relative to force magnitude in the vertical direction.

One result of this calculation was that the calculated TM joint reaction forces were higher or with similar values on the specimen balance side and with smaller values on the bolus side, which was interesting but still in conformity with [5]. The calculated *m. masseter* forces were higher on the bolus side, with a maximum value of about 150 N, while *m. temporalis* forces were higher on the balance side, with a maximum value of about 415 N, as shown in Fig. 3.

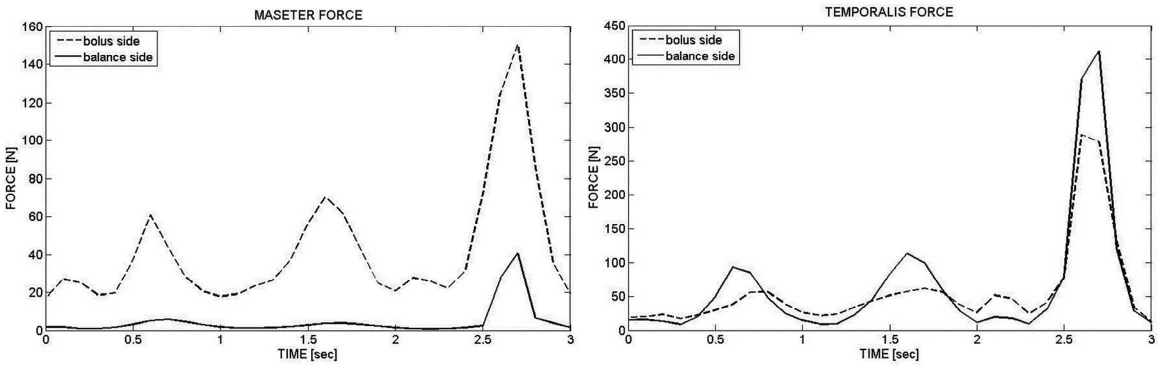

Fig. 3. Calculated *m. masseter* (left) and *m. temporalis* (right) forces on the bolus and balance sides with all muscles on both sides

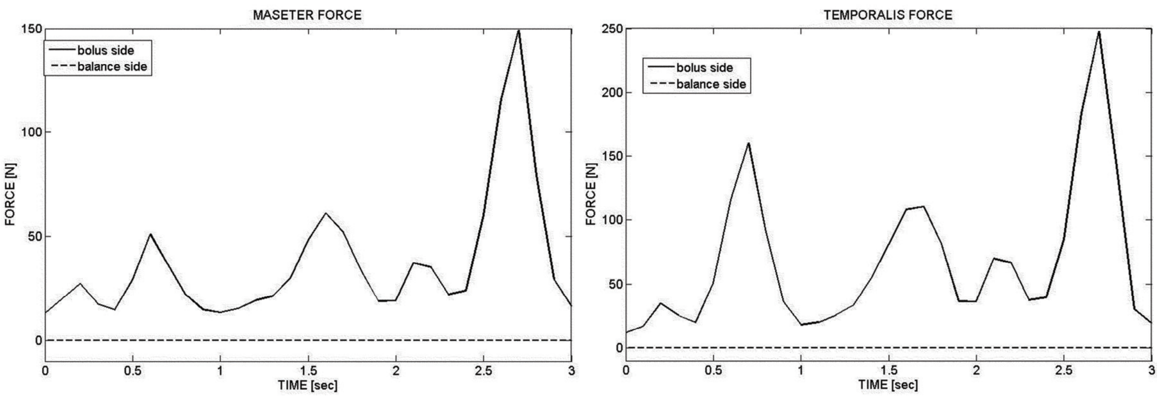

Fig. 4. Calculated *m. masseter* (left) and *m. temporalis* (right) forces on the bolus side.
Muscles on the balance side were resected during implantation of an artificial TM joint. Resected muscles have zero force

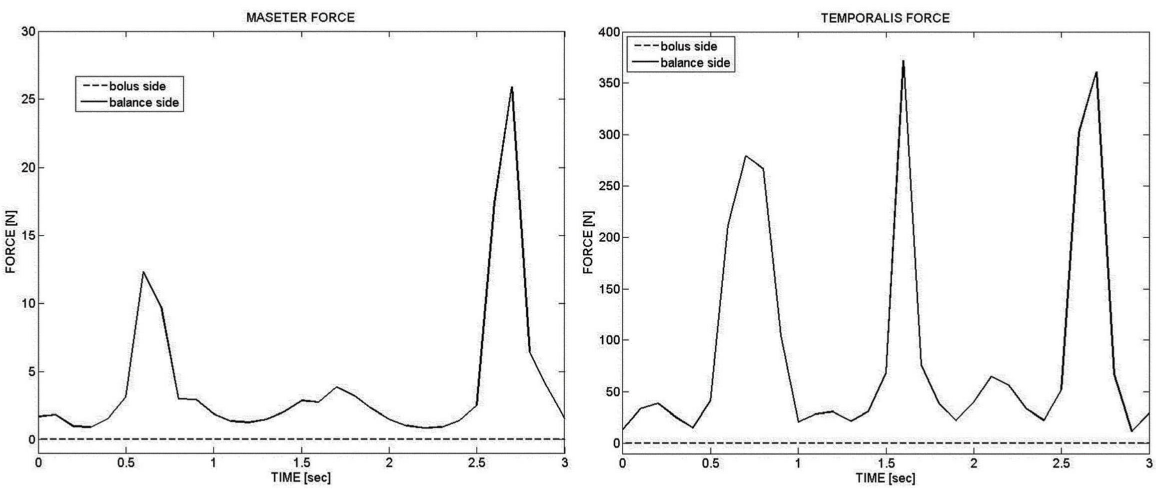

Fig. 5. Calculated *m. masseter* (left) and *m. temporalis* (right) forces on the balance side.
Muscles on the bolus side were interrupted during implantation of an artificial TM joint replacement.
Resected muscles have zero force

To represent resected *m. masseter* and *m. temporalis*, the force ratio was changed on both the bolus and balance sides. First, if the resected muscles are on the balance side, the effect of the *m. masseter* on bolus side had similar values to the case of muscles on both sides, but the effect of *m. temporalis* forces on the bolus side increases, as shown on graph in Fig. 4.

With regard to the effect of resected muscles on the bolus side, the *m. masseter* force on the balance side had similar values as muscles on both sides, but the *m. temporalis* force increased, as shown in Fig. 5.

Forces of the *m. pterygoideus medialis* in all the cases increased more on the side with the resected *m. masseter* and *m. temporalis*; which means that the *m. pterygoideus* has a role in closing the jaw. Other muscle forces associated with the resected *m. masseter* and *m. temporalis* were higher. Additionally, there was no significant difference between the bolus and balance sides, which shows their main role in stabilization. It is worth noting that the value of TM joint reactions forces were a little bit higher in the cases with resected muscles but similar on the bolus and balance side.

The results of the numerical simulations presented in [4] show a significant influence relative to the extent of muscle resection on contralateral TM joint loading (independent of bite location) in a total TM joint replacement. The biggest increase in the loading magnitude on the contralateral TM joint is most evident after *m. masseter* and *m. temporalis* resection (which is the most frequent method of total TMJ replacement in clinical practice). In the case of bite on molar M1 on the opposite side, this increase in loading is 8.4 times the physiological loading. Moreover, when comparing the results for individual types of resection it can be seen that in order to keep the best function of the TM joint on the opposite side after a total replacement, it is more important to save the *m. masseter* than the *m. temporalis*. If the *m. masseter* is saved, the total magnitude of loading on the TM joint for a bite on incisors I1 is 28% lower, and for a bite on molar M1 on the opposite side it is 127% lower.

4. Discussion

Our analyses used data from a 3D force micro sensor that measured masticatory forces. In our experiments the force sensor was implanted into the specimen on the bolus side. The force sensor was implanted in one tooth and masticatory force was measured only in one place. In reality masticatory forces are distributed among the surrounding teeth. Therefore this study has a limitation with regard to the measurement of the real magnitude of masticatory forces, however, this limitation would not influence the relationships observed in our results. We were not able to completely analyze the absolute value of applied forces, but we were able to study the relationships between them [17]. Inertia and mass properties of the mandible did not have a significant influence on results since mandible accelerations, during our investigation, were very small. TM joint forces observed during our investigation were smaller than published values [10] [11]. In our experiment a soft bolus (bread) was masticated, which probably explains why these forces were smaller than other published values.

Another potential limitation is that the actual magnitude of the forces of individual muscles was determined according to an accepted hypothesis that assumed masticatory forces are able to act independently in order to achieve the maximum bite force, and the resulting maximum force of a given muscle depends only on its physiological cross section, Koolstra et al. [5], [6]. The set-up of masticatory muscles is a general system of muscles, where there is no definite solution for muscle activity for a given operation. When verifying the mathematical model, Koolstra et al. [6] found no substantial difference between experimental measurements and the mathematical model.

In our analyses we represented the musculotendon complex as idealized mechanical objects, i.e., Hill type model. The disadvantages of these types of models include the assumptions associated with the input data, such as kinematic measurement of actuator length, and the muscle parameters of the model. The length of the muscle actuators, respectively, their origins and insertions were taken from literature. Their extensions and contractions were detected by experimental examination of chewing movements. The individual positions of each muscle insertion were not considered in the analyses.

The model for calculation of masticatory and TM muscle forces was assembled from the mandible and maxilla, including the skull, but did not include TM joint disks. Thus articulations between the upper and lower jaws were idealized joints. This simplification would affect the distribution of stress around the TM joint, for example, determined using FEM, but would have no effect on the resulting response in the TM joint set of balance of forces of chewing mechanism.

5. Conclusions

This study describes a 3D model that reflects the mechanism of the skull and lower jaw consisting of two bodies and two joints. Kinematical and force conditions used in this model were obtained experimentally, muscle physiological properties came from published literature. The aforementioned data were used for creating equilibrium equations and the problem was solved using static optimization, according to various criteria. The muscle strength and muscle forces calculated from the equations compared favorably with those presented in the literature [7], [11].

The calculation was done also with a simulation of a TM joint implant, which means the *m. masseter* and *m. temporalis* were interrupted where the TM joint was implanted. We demonstrated that all muscles on both the bolus and balance sides, had increased force values and helped in stabilizing jaw elevation. With regard to jaw elevation involving a resected *m. pterygoideus*, we found no resection differences associated with the muscles relative to the bolus or balance side.

The analyses showed that resection of the *m. temporalis* and especially of the *m. masseter*, which is the most common method of total TM joint replacement, had disastrous effects on contralateral joint loading. Conclusions of Tanaka et al. [13] and Koolstra et al. [8], together with the results of our FE analysis suggest that increased loading due to *m. masseter* and *m. temporalis* resection during TM joint replacement is the main cause of subsequent degeneration [3] of the contralateral TM disc and surrounding structures in what had been a healthy TM joint prior to surgery.

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