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ABSTRACTS 47th congress of the Society of Biomechanics

1. Introduction

Breast cancer (BC) is the most common disease affecting menopausal women worldwide. By estimation, 1.1 billion women are susceptible to be affected by 2025 (Ait Oumghar et al. 2020). BC tumour cells metastasise into bone sites and affect directly bone remodelling (BR) cells through their secretion of different cytokines. Those interactions may lead to osteolytic, osteoblastic and mixed lesions. Apart from the direct effect of BC on bone cells occurring after metastasis, primary tumour cells can also affect bone through the circulation of their secreted substances. According to (Chiou et al. 2021), those factors induce perturbed bone growth. In this work, we aim to represent the effects of primary BC tumour cells on bone cells within the bone microenvironment by means of a BR mathematical model. The primary BC has been represented by the effect of extracellular vesicles (EVs) released by tumour cells that affect distantly the receptor activator of nuclear factor (NF)- κ B Ligand (RANKL), the interleukin-6 (IL-6), and the osteoprotegerin (OPG) secretion of osteoblasts as studied in (Loftus et al. 2019), and by stimulating bone formation due to mineral apposition.

2. Methods

2.1. Bone remodelling model

The bone remodeling mathematical model of (Pivonka et al. 2013) has been considered for the modeling of primary BC effect on the bone growth, where preosteoblasts (OBp), active osteoblasts (OBa), and active osteoclasts (OCa) evolution is calculated over time. The proliferation of OBp is controlled by a mechanical stimulus driving by the strain energy density. As we are interested in menopausal women which suffer more frequently from BC, and giving that we will consider IL-6 effect on BR, we simulated normal conditions by incorporating estrogen, IL-6 in addition to Dickkopf (DKK-1), and Wnt Hill functions in the model in such a way they control

RANKL and OPG production by osteoblasts. The BR model parameters (Table 1) have been taken from (Wang et al. 2011) work, where the parameters are estimated based on experimental data from the literature. In order to represent normal conditions, we calibrated the differentiation rate of OCp D_{OCp} , the apoptosis rate of OBa A_{OBa} , and the bone formation rate k_{form} , to maintain unchanged BV/TV over time.

2.2. Primary breast cancer effect modelling

To mimic the primary BC effect on bone cells behavior, we have considered the EV action on the cells based on the results of (Loftus et al. 2019), which provided the effect of EV secreted by estrogen positive (ER+) BC cells MCF-7 on osteoblasts. The results showed that EVs induce an 8-fold and 2-fold increase of RANKL and IL-6, respectively, together with a 4-fold decrease of OPG production by osteoblasts. For the sake of modeling those observations, we have multiplied the functions $F_{EV,RANKL}$, $F_{EV,OPG}$ and $F_{EV,IL6}$ (Eqs. (2)–(4)) by the functions representing the concentration of RANKL, OPG and IL-6, respectively.

Each of $F_{EV,RANKL}$, $F_{EV,OPG}$ and $F_{EV,IL6}$ depends on BC cells' concentration in the breast environment C_T (Eq. (1)). P_T is the proliferation rate of BC cells, C_{Tmax} is the maximum concentration of BC cells, and $\pi_{act,E}^T$ is the estrogen Hill activation function that stimulate the ER + BC cells proliferation. $L_{EV,RANKL}$, $L_{EV,IL6}$, and $L_{EV,OPG}$ are, respectively, the constants driving of RANKL and IL-6 increase and the OPG decrease under EV's effect.

$$\frac{dC_T(t)}{dt} = P_T \pi_{act,E}^T \ln\left(\frac{C_{Tmax}}{C_T}\right) C_T \quad (1)$$

$$F_{EV,RANKL} = L_{EV,RANKL} C_T \quad (2)$$

$$F_{EV,OPG} = L_{EV,OPG} / C_T \quad (3)$$

$$F_{EV,IL6} = L_{EV,IL6} C_T \quad (4)$$

In addition to EVs action, we have mimicked the increase of mineral apposition by altering the rate of bone formation (Eq. (5)) affecting BV/TV (Eq. (6)).

Reliability and sensitivity of MMG and EMG signals during isometric contractions of upper limb muscles

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1. Introduction

Mechanomyography (MMG) is defined as the measurement of low-frequency lateral oscillations of active muscle fibers and is considered as the mechanical counterpart of electromyography (EMG) (Talib et al. 2018). MMG can be obtained through different means: microphones, lasers, piezoelectric crystals, and the most common in literature, accelerometers (Islam et al. 2013). MMG accelerometers are preferred over other methods because of their light weight, inexpensiveness, physical units (m/s^2), and relatively high signal-to-noise ratio (Islam et al. 2013; Talib et al. 2018). Compared to EMG, MMG is not sensitive to impedance, electrical interferences and less sensitive to placement, making it a suitable alternative (Talib et al. 2018). Despite such validation, only a few studies focused on polyarticular movements which could enable the use of MMG for ergonomic assessment of the workspace.

In this context, the present study is the first step aiming to compare the sensitivity and reliability of MMG signal vs. EMG during isometric contractions of the biceps brachii (BB), triceps (TRI), and deltoid (DEL) muscles for three force conditions.

2. Methods

2.1. Participants

Twenty-two male participants (age: 25.5 ± 5.3 ; height: 1.79 ± 0.07 m; weight: 75.7 ± 11.9 kg) participated in this study. Each participant was right-handed and was not subject to any musculoskeletal disorder in the last six months. Participants provided written informed consent before taking part in this experiment. This study was approved by the Academic Research Ethics Committee (Université Paris-Saclay, 2021-287).

2.2. Testing procedure

The testing procedure firstly consisted in measuring the maximum voluntary contraction (MVC) of studied muscles. The participant was asked to

Table 1. Reliability scores for each sensor, intensity level, and muscle.

Sensor	%MVC	25	50	75
EMG	Biceps	0.952	0.929	0.980
	Triceps	0.955	0.986	0.973
	Deltoid	0.960	0.960	0.966
MMG	Biceps	0.945	0.947	0.967
	Triceps	0.949	0.967	0.831
	Deltoid	0.958	0.935	0.863

Bold values are >0.9 .

perform a maximal contraction of three seconds against a force sensor screwed to an adjustable metallic structure for the MVC measurement. For the BB and the TRI muscle, the participant had the elbow flexed at 90° against the body and the force sensor placed on the wrist. For the DEL muscle, the arm was extended forward parallel to the ground with the force sensor on top of the wrist. The participant then performed submaximal contractions at 25, 50 and 75% of respective MVC. It consisted in maintaining the required force level for 15 seconds thanks to a force visual feedback on a screen. MMG and EMG signals were recorded during both maximal and submaximal contractions. Each MVC and submaximal contraction was repeated three times. A 2-minute and 1.5-minute rest period followed each MVC and submaximal contraction, respectively. The MVC value was the highest of the three trials. Testing order for force conditions was entirely randomized.

2.3. Sensors

The EMG signal was recorded using sensors (Miniwave, Cometa, Italy) placed following the SENIAM recommendations (Hermens et al. 1999) Force data were measured with sensor 2715-ISO, Sensy, Jumet, Belgium. EMG and force signals were sampled at 2000 Hz by default according to QTM software settings (Qualisys, Göteborg, Sweden). The MMG signal was recorded using triaxial accelerometers sensor (Moten Technologies, France) with a 1000 Hz sampling rate. Each sensor was placed on the muscle's belly following recommendations from the literature (Ibitoye et al. 2014).

2.4. Signal processing

EMG and MMG signals were band-pass filtered between 20 and 500 Hz (De Luca et al. 2010) and 5–100 Hz (Ibitoye et al. 2014), respectively. Each trial was segmented to keep only the middle 33% contraction time. The segmented signal amplitude was calculated using the Root Mean Square (RMS) and normalized based on MVC values.

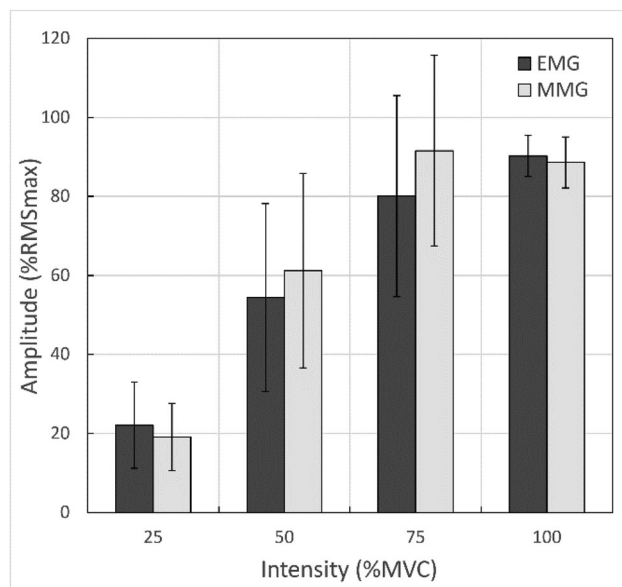


Figure 1. Mean normalized EMG and MMG RMS of the BB versus intensity for all participants.

2.5. Statistical analysis

EMG and MMG reliability were analyzed with an Intraclass Correlation Coefficient (ICC) (3, k). Two-way repeated-measures ANOVAs (muscle \times intensity) were performed separately for RMS_{EMG} , and RMS_{MMG} . Pairwise comparisons were conducted using the Bonferroni correction (JASP version 0.15).

3. Results and discussion

The EMG and MMG recordings exhibited reliability scores >0.9 overall, with ICC scores of 0.96 ± 0.02 and 0.93 ± 0.05 , respectively (see Table 1).

ANOVAs exhibited significant main effects of intensity on RMS amplitude. For both sensors and each muscle, mean RMS amplitudes were statistically different from each other ($p < 0.001$), except between 75% MVC and 100% MVC (see Figure 1).

Overall MMG reliability scores obtained in this study are above previous results (ICC = 0.79) obtained with a microphone MMG (Meagher et al. 2020). The increase of RMS_{EMG} and RMS_{MMG} as a function of intensity agrees with previous studies (Ibitoye et al. 2014) and can be related to the increasing number of active motor units (MU) from 25% to 75% MVC. Around 75% MVC, the majority of MU is recruited, causing an amplitude plateau. To achieve 100% MVC, there is an increase of MU firing rate. This increase tends to maintain or even decrease the amplitude above 75% MVC (Islam et al. 2013; Talib et al. 2018).

4. Conclusions

The present study shows that MMG sensors provide equivalent reliability and sensitivity to EMG sensors during isometric contractions. Results from this preliminary study agree with the literature, which permit to consider new developments for ergonomic challenges. Further work is now needed on dynamic contractions, which have been less studied in the literature, to assess the feasibility of MMG recordings for in-field applications.

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Disclosure statement

No potential conflict of interest was reported by the authors.

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