Summary

Muscle load sharing, an energy-based approach
Musculoskeletal models are a valuable tool in the study on human movement. When the kinematics and external forces that act on the human body are known, such models can be used to calculate the resultant joint moments for a given posture or motion by simple Newtonian mechanics. It is, however, difficult to determine the contribution of the individual muscles to these moments. In general, there are more muscles crossing a joint than is theoretically necessary in order to perform all possible movements. This is called the indeterminacy problem or the load sharing problem. Inverse dynamic models often make use of cost functions to solve this load sharing problem. The use of cost functions is based on the assumption that the central nervous system controls the musculoskeletal system in an optimal manner, optimising a certain cost or objective. It is, however, difficult to find the right criterion, since it is unknown what quantity is optimised in real life. Therefore, only assumptions can be made.
Most cost functions, that have been proposed, are mechanical cost functions, which are based on muscle force and often weighed by maximal force or a morphological parameter such as physiological cross sectional area (PCSA). Although some of these functions are assumed to be related to physiological costs like energy consumption or fatigue, clear relationships have never been proven. The validation of cost functions is hampered by the fact that muscle force can not easily be measured in vivo and is therefore often restricted to a comparison between calculated muscle force and recorded EMG patterns. Especially for submaximal activities, it has often been assumed that movements are performed by minimising energy consumption. Nevertheless, up till now there are no cost functions defined, based on muscle energy consumption. The aims of this thesis were to define a cost function that represents muscle energy consumption and to validate this cost function with a metabolic parameter.
In a model, the predicted function of a muscle depends on the total number of degrees of freedom included. Not only the number of degrees of freedom, used for a single joint, is of great influence for muscle function. Because of the special role of bi- and poly-articular muscles, it also depends on the distribution of muscle forces around adjacent joints. As a consequence, the effect of cost functions on model predictions should ideally be studied with the help of models that include all degrees of freedom instead of using simple one- or two joint models. Therefore, in the current thesis the Delft Shoulder and Elbow...
Model (DSEM) was used, which is a 3D inverse dynamic musculoskeletal model of the upper extremity. Validation of the cost functions, studied in this thesis, was done with the help of Near InfraRed Spectroscopy (NIRS). NIRS is a non-invasive method which can be used to measure the oxygenation of biological tissue. It has been proven to be a valid method to measure muscle oxygenation. Up till now, however, little is known about the relationship between muscle oxygen consumption (\(\dot{V}O_2\)), determined with NIRS, and muscle activation, determined with EMG.

When a muscle is stimulated, calcium is released from the sarcoplasmatic reticulum (SR), enabling the cross-bridges to attach and consequently to produce force. The two major energy-consuming processes in a muscle are the re-uptake of calcium in the SR (activation dynamics) and the detachment of cross-bridges (contraction dynamics). Since EMG and NIRS apply to different levels of the process of force production, it is plausible that these two techniques show different results. NIRS registers muscle oxygen consumption and hence gives an indication of both the activation and the contraction dynamics process. EMG, on the other hand, registers the excitation of the muscle only, which initiates the calcium flow from the SR and therefore should give an indication of the contraction dynamics only.

The exact relationship between the EMG signal and calcium flow or cross-bridge attachment is unknown. Since activation is not linearly related to force, we expected a non-linear relationship between the EMG and \(\dot{V}O_2\). In Chapter 2 we performed experiments to investigate this relationship. Both EMG and \(\dot{V}O_2\) of two arm muscles (m. biceps brachii caput breve and m. brachioradialis) were measured during several isometric contractions in which subjects had to perform combinations of elbow flexion and pro/supination moments at force levels up to 70% of their maximum force. The results of these experiments showed that both EMG and \(\dot{V}O_2\) were linearly related to the performed external moment and that the relationship between \(\dot{V}O_2\) and EMG could be described by a linear equation.

In Chapter 3 an experiment was performed in which the \(\dot{V}O_2\) of four elbow muscles (the m. biceps brachii caput breve, the m. biceps brachii caput longum, the m. brachioradialis and the m. triceps brachii caput laterale) was recorded. The subjects had to perform isometric contractions, generating several combinations of elbow flexion/extension and pro/supination moments. The measured \(\dot{V}O_2\) of the arm muscles was used in this experiment to validate two different cost functions. The first cost function was the well known stress cost function, in which the sum of squared muscle stress is minimised. Although it has been suggested previously that the stress cost function
is related to energy consumption, validation with a metabolic variable was never done. The second cost function, used in this study, was a newly proposed cost function which is further referred to as the energy-related cost function. This cost function is based on the two main energy-consuming processes of the muscle mentioned before, namely activation dynamics and contraction dynamics. The metabolic cost of a muscle is equal to the summed energy cost over the activated sarcomeres. Therefore, not only the PCSA, representing the cross-bridges that are in parallel, but also muscle fibre length, which is related to the number of sarcomeres in series, should be included in the cost function. For that reason, in Chapter 3 both PCSA and fibre length were incorporated in the new cost function. It was expected that this energy-related cost function would give a better representation of muscle energy consumption than the stress cost function. To find out if this was really the case, both cost functions were implemented into the inverse dynamic version of the Delft Shoulder and Elbow Model. For each of the four measured elbow muscles experimentally obtained \( \dot{V}O_2 \) values were compared to individual predicted cost values. The analyses focused on two different measures: first the overall fit (correlation and RMSE values) between model predictions and experimental results and secondly the number of the so-called ‘false negatives’ and ‘false positives’ (conditions in which the model predicted NO muscle activity where muscle activity was indeed recorded, and vice versa). For the m. triceps brachii caput laterale the stress cost function led to a good correspondence between \( \dot{V}O_2 \) and cost. For the flexor muscles, however, the predicted metabolic cost was significantly lower than the experimentally obtained \( \dot{V}O_2 \) values and a large number of ‘false negatives’ were found. It seemed that, compared to flexion/extension moments, pro/supination moments had a disproportional large effect on the predicted activity of these muscles. Compared to the stress cost function, the energy-related cost function showed a better correspondence between \( \dot{V}O_2 \) and cost and fewer ‘false negatives’. It was therefore concluded that the energy-related cost function appears to be a better measure for muscle energy consumption than the stress cost function and leads to more realistic predictions of load sharing.

Although the measurements, described in Chapter 3, included several combinations of flexion/extension and pro/supination moments, not all possible combinations of joint moments were included. Furthermore, all measurements were performed at one fixed elbow angle (90°) only. Consequently, muscle length was not varied, which means that the force-length characteristics were not taken into account. Therefore, in Chapter 4 new experiments were performed in which elbow angle (and therefore muscle length) was varied for a range of elbow flexion angles.
from 55° to 120°. Two different experiments were performed in Chapter 4. In the first experiment, EMG of eight elbow muscles was recorded while the subjects (n = 6) performed a full range of all possible (49) combinations of elbow flexion/extension and forearm pro/supination moments, varied over three different force levels. All 49 measurements were repeated at four different elbow angles. The set-up of the second experiment was comparable to the first experiment; except that in this experiment besides EMG measurements VO₂ measurements with NIRS were performed as well. Due to the fact that NIRS measurements are very time-consuming, the number of moment combinations, muscles and subjects was limited compared to the first experiment.

When the elbow angle changes, not only the lengths of the muscles, which span the elbow joint, change but the moment arms as well. Since the maximal force, that can be produced by a muscle, depends on its actual muscle length and the moment that can be produced depends on the actual moment arm, it seems obvious that elbow angle will influence muscle activity. It is, however, unclear what the effect of this change in elbow angle will be on the muscle load sharing. From a single muscle perspective it could be expected that muscle activity decreases at optimum length/optimum moment arm as the muscle can achieve the same force/moment with less activity. Conversely, from a multiple-muscle point of view the opposite can be expected: as the muscle becomes ‘cheaper’ its contribution might increase, since the use of that muscle has become ‘cheaper’.

The results of Chapter 4 showed that joint angle, and therefore moment arm and muscle length, influences both the activity level of the muscle as well as the load sharing between muscles. Unfortunately, the principles behind this load sharing were difficult to quantify, since it was impossible to distinguish all the individual aspects that affect muscle activity. Furthermore, the results of Chapter 4 confirmed the conclusion of Chapter 2 that the relationship between EMG and VO₂ can be described as a linear relationship and that this relationship is not influenced by elbow angle. Another important result of Chapter 4 was that although, in general, subjects showed comparable muscle activation patterns, there were also some striking inter-individual differences. Assuming that the subjects used the same optimisation strategy, these inter-individual differences might be explained by differences in muscle morphology. As biomechanical models are in general based on a single anthropometrical data set, inter-individual differences will not be reflected by these models.

In Chapter 5, it was investigated whether the correspondence between model predictions and recorded muscle oxygen consumption (presented in Chapter 4) could be improved when for each subject an individualised
morphological data set was used. If subjects indeed differ in muscle morphology, it is possible that for any given subject a different combination of for instance PCSA and moment arm values might lead to a better fit between experimental and model data. In Chapter 5, also the interaction between morphology and optimisation criteria was investigated. Again, model simulations were done with the DSEM and the stress cost function was compared to the energy-related cost function. In this experiment a new morphological data set was used in the model. This new set includes data on optimal fibre length, which made it possible to implement force-length characteristics. The energy-related cost function could therefore be reformulated such that the force-length characteristics of the muscles were included. The PCSA and moment arm values of eight upper- and forearm muscles were varied in the model. An approximative optimisation strategy was used to find for each subject the individual parameter set that was expected to correspond best with the experimental data. The fit between the experimental data and the model predictions, done with this optimal morphological parameter set, can be seen as the maximum improvement that can be attained by the inclusion of individual morphology in the model.

The results showed that the load sharing is strongly dependent on cost function, as well as on morphology. The modelling results improved by using the individually optimised morphological data. This could explain part of the inter-individual variability in the experimental results, and also led to a better morphological parameter set for individual subjects than the ‘generic’ model morphology that was based on one specimen only. Fitting individual morphological data, however, was only effective for the energy-related cost function. Although not all the ‘false negatives’ disappeared, the energy-related cost function again showed a better fit to experimental results than the stress cost function. The remaining ‘false negatives’ were predominately found in conditions, including a moment for which the particular muscle had no or even the opposite effect. In these conditions, muscles seem to be activated due to a compensating effect for one of the other muscles or due to their effect on another degree of freedom.

In Chapter 6, the Epilogue, the main findings of the research, described in this thesis, were summarised and discussed. The present thesis shows that the use of Near Infrared Spectroscopy, to measure muscle oxygen consumption is a valuable development in cost function validation. Furthermore, it becomes clear that load sharing, not always leads to the same muscle activation pattern and that it is influenced by different factors. As the exact principles behind load sharing are still unknown, it is difficult to find the right optimisation criterion (cost function). In
addition, the evaluation of a cost function is hampered by the fact that the predicted load sharing is not only influenced by the type of cost function but also by factors such as morphological parameters. It is difficult, if not impossible, to separate the individual effects of morphological parameters on the predicted load sharing. In Chapter 6, it is also stressed that validation of cost functions can only be done with multiple-muscle and -joint models that include the actual number of degrees of freedom as well as all actual adjacent joints.

It is clear that the stress cost function does not predict realistic muscle activation patterns, since using this cost function resulted in a relatively large amount of ‘false negatives’ and ‘false positives’. The energy-related cost function performed better, but also needs improvement. The fact that ‘false negatives’ and ‘false positives’ were mainly seen in paradoxical conditions in which the activated muscle contributes to one of the requested moments only and counteracts to the other moment or in which the muscle does not directly contribute to one of the requested moments at all emphasises the significance of including a wide range of force tasks into the experimental set-up, covering all degrees of freedom.

Since the energy-related cost function still shows some discrepancies with experimental data, it can be questioned whether it is indeed energy consumption that is optimised during submaximal movements. Furthermore, it can be argued whether the energy-related cost function, proposed in the present thesis, is a valid description of muscle energy consumption. Further improvement of the cost function can possibly be achieved by optimising the weight factors of the cost function. It should also be studied whether energy consumption is optimised on the level of the individual muscles, independent of and not necessarily leading to a minimisation of overall energy consumption. Based on the observed inter-individual differences in the present thesis, it can also be questioned if the assumption of a general optimisation criterion is valid and that people use different optimisation strategies. The results of Chapter 5 indicated, however, that part of the inter-individual differences could be explained by possible differences in morphology, although this was only effective using the energy-related cost function. Inter-individual differences could, on the other hand, also be caused by different strategies in coping with stability. It is likely, that proper validation of cost functions can not be done without including a task constraint such as stability.